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Schneider

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[54] POLYLACTIDE SUTURES

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Related U.S. Application Data

[63] Continuation-in-part of Ser. No. 700,036, Jan. 24, 1968, abandoned, which is a continuation-in-part of Ser. No. 449,630, Apr. 20, 1965, abandoned, which is a continuation-in-part of Ser. No. 308,685, Sept. 13, 1963, abandoned, which is a continuation-in-part of Ser. No. 231,860, Oct. 19, 1962, abandoned.

[52] U.S. Cl. 128/335.5, 260/78.3

[51] Int. Cl. A61l 17/00

[58] Field of Search 128/334, 335.5; 260/78.3

[56] **References Cited**

UNITED STATES PATENTS

2,668,162	2/1954	Lowe	260/78.3
2,703,316	3/1955	Schneider	260/78.3
3,225,766	12/1965	Baptist et al.	128/335.5
3,297,033	1/1967	Schmitt et al.	128/335.5

FOREIGN PATENTS OR APPLICATIONS

5,354	11/1964	Iran	128/335.5
779,291	7/1957	Great Britain	260/78.3

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Attorney—Robert W. Kell and Robert L. Minier

[57] **ABSTRACT**

Absorbable surgical sutures that are dimensionally stable within the body may be prepared by the extrusion of polylactide polymer, including copolymers of L(−) lactide with up to 35 mole percent of glycolide. Said polymers are characterized by an inherent viscosity of at least 1.0, and the extruded filaments are oriented by drawing at a temperature of about 50° to about 140° at a draw ratio of up to 11x, and annealed. Sutures so prepared have a tensile strength of from 25,000 p.s.i. to 100,000 p.s.i.

79 Claims, 5 Drawing Figures

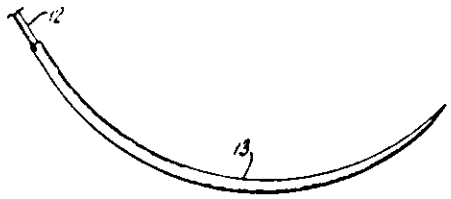

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Fig. 1.

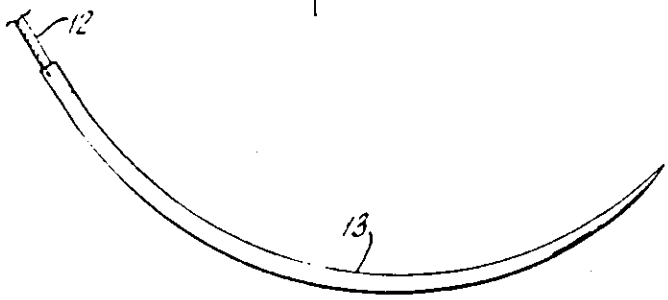


Fig. 2.

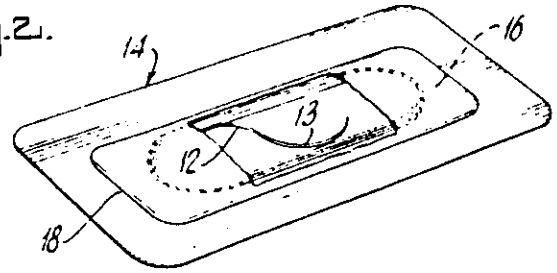


Fig. 3.

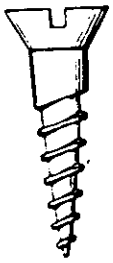


Fig. 4.

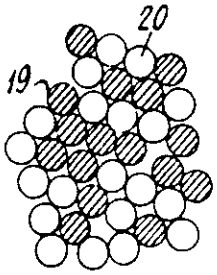
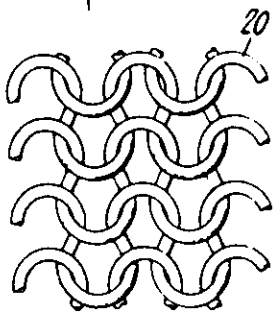


Fig. 5.



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POLYLACTIDE SUTURES

This application is a continuation-in-part of my copending U.S. application Ser. No. 700,036, filed Jan. 24, 1968, now abandoned, which in turn was a continuation-in-part of my then copending U.S. application Ser. No. 449,630, filed Apr. 20, 1965, now abandoned, which in turn was a continuation-in-part of my then copending U.S. application Ser. No. 308,688, filed Sept. 13, 1963, now abandoned, which in turn was a continuation-in-part of my then copending U.S. application Ser. No. 231,860, filed Oct. 19, 1962, also now abandoned.

This invention relates to new articles of manufacture and to their use. More particularly, the invention is concerned with surgical aids prepared from synthetic polymers including copolymers of lactic acids and their use in surgical applications, e.g., sutures and ligatures and other prosthetic devices used in joining or supporting living tissues.

Catgut (actually from sheep or beef intestine) is the most commonly used absorbable suture now on the market. In many instances, however, it may cause adverse tissue reaction in the sutured flesh. This, together with the fact that it requires storage under moist conditions, makes it less than an ideal suture material. Nylon, stainless steel, cotton, linen, ramie, "Teflon" fluorocarbon resin, "Dacron" polyester fibers, silk, and other materials have been suggested and/or used as surgical sutures. Some of them have advantages over catgut in strength, uniformity, and storage characteristics, but they are not absorbed by living tissue.

Among the requirements of the ideal absorbable suture product are that it should handle properly, should approximate and hold tissue for proper healing with the least possible damage, should not tear tissue, should have adequate tensile strength, should be controllably uniform in properties, including dimensional stability within the body, should be sterilizable, should be absorbable by living tissue, preferably at a constant rate regardless of the place in the body and the condition of the patient, without causing such unfavorable tissue reactions as walling off, granuloma formation, excessive edema, etc., and finally should be capable of tying and holding surgical knots properly.

This invention fulfills the above requirements to a remarkable degree by providing highly oriented, high tenacity filaments of polymers and copolymers of lactic acid, the filaments having excellent dimensional stability in body tissue and preferably retracting less than 10 percent in an empirical test in which the filaments are immersed in water at 37° C. for a period of 24 hours.

These filaments are prepared from lactic acid homopolymers and copolymers having an inherent viscosity of at least 1, preferably above 1.2, as determined at 0.1 percent concentration in benzene by weight at 25° C. prior to being oriented. Any polylactide composition containing up to about 15 percent by weight of repeating units of the formula:

(1)

wherein R is lower alkylene, preferably methylene ($-CH_2-$) or ethylene ($-CH_2CH_2-$), m is 0 or 1, R' is hydrogen or lower alkyl, R'' is hydrogen or alkyl of up to about 22 carbons when m is 0 and hydrogen or lower alkyl when m is 1, and can be the same as R' or different, can be employed to make the sutures of this invention. Preferred, because of availability of starting materials, are repeating units derived from alpha-hydroxycarboxylic acids, i.e., units of the above formula in which m is 0. Most preferred, because of the properties of the sutures made therefrom, are repeating or comonomer units derived from glycolide or DL-lactide, i.e., repeating units of formula (1) in which m is 0, R' is hydrogen or methyl, and R'' is hydrogen. In other words, the number of carbon atoms in the repeating unit is two to about 24, preferably two to about eight, and most preferably two to three. It will be understood

that when m is 0, R' is methyl, and R'' is hydrogen, the repeating unit in formula (1) could be derived from DL-lactide. This would result in a copolymer containing both antipodal species derived from alpha-hydroxypropionic acid. When the repeating unit in formula (1) is identical with the principal unit, the polylactide composition is a homopolymer. In the specific instance when m is 0 and both R' and R'' are hydrogen, (when glycolide is the comonomer), the polylactide composition may contain about 35 mole percent of repeating units of the formula

(2)

Such copolymers of L(−) lactide and glycolide may also be employed to make the sutures of this invention.

Illustrative of the comonomers which can be employed with the lactide to form copolymers useful in preparing the filaments of this invention, there can be name glycolide, beta-propiolactone, tetramethylglycolide, beta-butyrolactone, gamma-butyrolactone, pivalolactone, and intermolecular cyclic esters of alpha-hydroxybutyric acid, alpha-hydroxyisobutyric acid, alpha-hydroxyvaleric acid, alpha-hydroxyisovaleric acid, alpha-hydroxycaproic acid, alpha-hydroxy-alpha-ethylbutyric acid, alpha-hydroxyisocaproic acid, alpha-hydroxy-beta-methylvaleric acid, alpha-hydroxyheptanoic acid, alpha-hydroxyoctanoic acid, alpha-hydroxydecanoic acid, alpha-hydroxymyristic acid, alpha-hydroxystearic acid, and alpha-hydroxylignoceric acid.

The filaments prepared from the above-described lactide polymers and copolymers are conveniently formed by melt-extruding the polylactic acid through a spinneret and then drawing the filaments in one or more stages to about four times their original length to effect orientation and to improve their tensile strength. The resultant oriented filaments are strong and retain much of their strength on being tied into surgeon's knots.

To further improve their dimensional stability and particularly tensile strength retention, one may subject them to an annealing treatment. This optional annealing treatment is effected by heating the filament, while holding it essentially taut, at 60° to 150° C., and then allowing it to cool to room temperature (25° C.) while held taut. The annealing is preferably conducted for such a time that the filament shows less than 10 percent shrinkage on subsequent immersion, for 24 hours without tension, in water at 37° C. The heating step of annealing usually requires from 0.5–5 minutes, to as long as 1 week.

A filament which meets the foregoing shrinkage test (37° C.) undergoes substantially no shrinkage when used as a suture in contact with body tissues (see example II). The conditions of this test are designed to give a quick in vitro measure of the dimensional stability of the filaments that can be projected to their usefulness as suture materials. In this connection, it should be mentioned that the conditions of draw have an influence over the shrinkage. Further, it has been found that those filaments showing little shrinkage in 24 hours at 37° C. have relatively little shrinkage when implanted in an animal body.

Since the function of a suture is to join and hold severed tissue until healing is well along, and to prevent separation as a result of movement or exercise, the suture should have adequate strength. It is particularly important that strength be maintained when knots are tied and during the actual procedure of drawing tight a suitable knot. Filaments from lactic acid polymers in high molecular weight oriented form are exceptionally strong and most significantly retain a high proportion of their strength at the knot point, as shown in the following table.

TABLE I

	Tensile strength (straight pull) p.s.i.	Percent elongation at break	Tensile strength (surgeon's knot) p.s.i.*	Percent loss in strength, knot vs. straight
Poly-L(−) lactide	[1] 130,000	17	83,000	29
Do	[2] 100,000	16	78,000	28
Catgut	[3] 80,000	20	29,000	42
	[4] 44,000	20	27,000	39
L(−) lactide/gamma-butyrolactone (95/5) copolymer	[5] 59,000	17	42,000	29

[1] Inherent viscosity = 2.8, 10X draw, 0.006 inch diameter.
[2] Inherent viscosity = 2.8, 10X draw, 0.008 inch diameter.
[3] Chromic gut (0.006–0.010 inch diameter).
[4] Chromic gut (0.010–0.013 inch diameter).
[5] After U.S. Pharmacopoeia
[6] Inherent viscosity (bulk polymer) = 3.0 (spun filament) = 1.6, 10X draw, 0.009 inch diameter.

As will be apparent from table I, the inherent viscosity of the spun filament, i.e., the oriented filament, may be somewhat less than that of the bulk polymer or copolymer, for during the extrusion operation some degradation of the polymer may occur depending on the extrusion conditions employed. If the sutures are sterilized by high energy radiation, there may be a further lowering of the molecular weight of the polymer, and a resulting decrease in tensile strength. However, by starting with lactide polymers and copolymers having inherent viscosities of at least 1, the sutures prepared therefrom are entirely satisfactory if one minimizes degradation during sterilization, even though there may be some loss in inherent viscosity due to extrusion and orientation.

The filaments of this invention are further characterized by their hydrolysis behavior and absorbability. On treatment with boiling water for 100 hours, they lose at least 20, and preferably at least about 50 percent, of their weight. On treatment with boiling water for a period of 50 hours, the copolymers lose at least about 8 percent of their weight, and preferably they lose at least about 35 percent of their weight.

By varying the type and amount of comonomer employed, the rate of hydrolysis (absorption) of the suture can be controlled. In contrast to the highly variable absorption rates of catgut, the absorption of polylactide polymers is relatively more independent of the place in the body where used and of the condition of the patient. Since the hydrolysis rate of a particular lactic acid polymer is constant at a fixed temperature, say, at 37° C., absorption can be speeded up, for instance, by using different copolymers. For example, poly-L-lactide was 15.3 percent absorbed in the back muscle of a rat after 270 days. Under comparable conditions, L(−)-lactide/DL-lactide (97/3) copolymer was 18.5 percent absorbed, L(−)-lactide/DL-lactide (95/5) copolymer was 29.0 percent absorbed, L(−)-lactide/glycolide (95/5) copolymer was 27.3 percent absorbed, and chromed catgut was 67 percent absorbed. The rate of absorption of a copolymer of L(−)-lactide and glycolide increase with increasing amounts of glycolide in the polymer chain.

As already indicated, high tensile strength is an exceedingly desirable characteristic for suture materials. The filaments of the present invention are characterized by having a tensile strength of at least 25,000 p.s.i., preferably above 40,000 p.s.i. Some have tensile strengths ranging up to 100,000 p.s.i. and higher. Their knot strengths, expressed in lbs. of pull, exceed the minimum limits set for absorbable sutures by the U.S. Pharmacopoeia, i.e., from 0.125 lb. for a 0.001–0.002 inch filament to 25 lbs. for a 0.036–0.040 inch filament.

In preparing the polymers and copolymers from which the filaments of this invention are made, the appropriate intermolecular cyclic ester or intramolecular cyclic ester (lactone) of the hydroxy acid is employed. These can be derived from pure D(−)- or L(+)-lactic acids, the optically inactive DL-lactic acid mixture, any desired mixtures of pure D(−)-lactic and L(+)-lactic acids, and other alpha, beta, or gamma-hydroxy acids, about which more will be said later. In general, it is preferred, for the preparation of lactic acid homopolymers and for the introduction of lactide repeating units into copolymers to use as a starting material a lactide derived from either the pure L(+)-acid or pure D(−)-acid because the polymers obtained therefrom have a higher melting point than those derived from the DL-acid mixtures, are much less water sensitive, are stronger, and have a greater degree of crystallinity. For example, the polylactides from the DL-acid melt at 130° to 140° C., whereas those from the L(+)-acid melt at 145° to 175° C. The polylactides from the L(+)-acid or D(−)-acid are less sensitive to alcohol, a commonly used disinfecting medium in surgery, than those from the DL-acid. The L(+)-form is more readily available than the D(−)-acid and hence is particularly preferred. It is to be understood that the various lactides can be made from the corresponding lactic acids by a variety of published methods including that described in Schneider U.S. Pat. No. 2,703,316.

Table II, below, summarizes data comparing the properties of polymers prepared from L(−) lactide with those prepared from DL-lactide.

TABLE II

	Polymer From	
	DL-lactide	L(−) lactide
inherent viscosity	0.7–2.0	0.7–3.5
melting point	130°–140° C	145°–175° C
optical activity	no	yes (−156°)
solubility	$CHCl_3$, benzene, acetone	$CHCl_3$, benzene, acetone
density	1.26	1.26
tensile strength at break (monofilament)	20,000–40,000 p.s.i.	70,000–100,000 p.s.i.
elongation at break (monofilament)	15–30 percent	15–30 percent
tensile strength at break (dry film)	26,000 p.s.i.*	29,000 p.s.i.*
inherent viscosity (film)	1.29*	1.27*
elongation at break (film)	48 percent*	23 percent*

*Taken from U.S. Pat. No. 2,758,987.

In general, the tensile modulus, melting point, and specific rotation of a lactic acid polymer is maximum for the homopolymer of a single-antipodal species and decreases with increasing amounts of the other antipodal species in the polymer chain. This characteristic of lactic acid copolymers is an advantage since it permits one to choose a copolymer composition that can be extruded to form filaments which have improved flexibility, without appreciable sacrifice in strength.

In preparing copolymers, the repeating units derived from comonomers discussed above are introduced by use of the appropriate cyclic esters. For repeating units derived from alpha-hydroxy acids, these are usually the intermolecular cyclic esters containing six-membered rings, e.g., glycolide. For repeating units derived from beta- or gamma-hydroxy acids, the monomeric lactones, e.g., beta-propiolactone and gamma-butyrolactone, are usually used.

The polymer filaments of the present invention may be woven, braided, or knitted either alone or in combination with nonabsorbable fibers such as nylon, polypropylene, ORLON, DACRON, or TEFLON to form tubular structures having use in the surgical repair of arteries, veins, ducts, esophagi and the like. The manufacture of such tubular structures wherein the wall of the tube is fabricated of absorbable and nonabsorbable threads is described in U.S. Pat. Nos. 3,304,557, 3,108,357, and 3,463,158, the teachings of which are incorporated herein by reference. Inasmuch as the polylactide filaments are thermoplastic such tubular grafts may be crimped on a mandrel at elevated temperature and upon cooling to room temperature, will retain the crimp.

Tubular structures of polylactide filaments may be prepared that are resistant to radial compression and expansion by applying a helical wrapping of polypropylene monofilament around the external surface of the tube and fusing the polypropylene to unite the helical wrapping with the polylactide filaments in the external surface of the tube as illustrated in U.S. Pat. No. 3,479,670.

The polymers of the present invention are also useful in the manufacture of cast films and other solid surgical aids such as scleral buckling prostheses. Thus, cylindrical pins, screws, reinforcing plates, etc., may be machined from the cast polymer having in vivo absorption characteristics depending upon the polymer composition and molecular weight.

The invention will appear more clearly from the following detailed description when taken in connection with the accompanying drawings which show by way of example preferred embodiments of the inventive idea. Referring now to the drawings

FIG. 1 is a perspective view of a needle-suture combination.

FIG. 2 is a perspective view of a suture-needle combination within a hermetically sealed container;

FIG. 3 illustrates a screw machined from the polymer of the present invention;

FIG. 4 is a cross-sectional view of a composite yarn containing filaments of different composition and,

FIG. 5 is a plan view of a knitted fabric.

In preparing the filaments of this invention, it is essential to use polymers made from highly-purified lactides. For example, for excellent results L(−) lactide should have a melting point of at least 96° C. and a specific rotation greater than −295°. The polymerization is effected by heating the lactide above its melting point, but below about 215° C. in the presence of a polyvalent metal oxide or compound thereof, under anhydrous conditions in an inert atmosphere.

Specially useful catalysts are zinc oxide, zinc carbonate, basic zinc carbonate, diethylzinc, titanium, magnesium or barium compounds, litharge, stannous octoate and the like.

The amount and type of catalyst used determine the particular temperature and time required to produce polymer useful for conversion to the filaments of this invention. Thus, the amount can be as low as 0.001 weight percent or as high as 2 weight percent. As a rule, the lower the amount of catalyst, the longer the time required to produce polymer of a given inherent viscosity and, conversely, the higher the catalyst concentration, the shorter the time. The best balance is usually obtained employing from 0.02 weight percent to 1 weight percent of catalyst.

In general, it is desirable to agitate the reaction mixture continuously during the polymerization in order to produce a homogeneous polymer at good conversions and to conduct the reaction in two steps, the first being carried out at a lower temperature than the second, or finishing step. Other methods, such as those disclosed in U.S. Pat. Nos. 2,703,316 and 2,758,987 can be used in making the polymers.

The following is a brief description of a method for preparing the polymer useful for conversion to the filaments of this invention: lactide, purified by several crystallizations from carbon tetrachloride, is placed with one or more solid comonomers in a thoroughly dried reactor equipped with a stirring bar, nitrogen inlet tube, and a drying tube filled conveniently with anhydrous magnesium sulfate or calcium chloride. Nitrogen, which has been dried by passage through anhydrous magnesium sulfate or calcium chloride, is introduced immediately above the reaction mixture and heating and stirring are started. When the temperature of the reaction mixture has reached about 100° C., the nitrogen inlet is replaced by a thermometer, and from about 0.001 to 2 weight percent of an oxide or salt, of group II metal of atomic number 12 through 56, or litharge is added. In the case of copolymerization with a liquid comonomer, the liquid comonomer is preferably added after the lactide has melted. Heating is continued until polymer having an inherent viscosity of at least 1 at 0.1 percent concentration in benzene at 25° C. is obtained. This may require from a few minutes up to 25 or more hours, depending upon the catalyst used.

Polymer, produced as above, may be suitably further treated by cutting it into small pieces, dissolving in a suitable solvent, for example, benzene, toluene, or xylene, and the polymer precipitated by pouring the solution into a large volume of a nonsolvent for the polymer, desirably hexane. The precipitated polymer is removed by filtration, transferred to a blender and a nonsolvent for the polymer is added. The blender is started and after a homogeneous mixture has been obtained, the mixture is filtered. The polymer is allowed to dry on the filter and is then transferred to a vacuum oven. After drying overnight at 100° C., the polymer is removed from the oven and allowed to cool to ambient temperature.

As already indicated, the polymer material can be converted to filaments by melt-extrusion and also by spinning from solution. The diameter of the resulting filaments may be as small as 0.001 inch or less for the individual strands making up the multifilament structures and as large as 0.045 inch for very heavy monofilament sutures. Generally, however, the filaments of this invention will not have a diameter greater than 0.020–0.025 inch. Preferred are monofilaments having diameters of about 0.001–0.020 inch and multifilament structures having individual filaments of from less than 0.00025 to 0.003 inch diameter.

It will be understood that spinning and drawing may be done singly or in multiples. To prepare multifilament braided sutures, one may take either monofilaments or groups of filaments to braid.

Spinnerets having orifice sizes of 0.005 inch or larger, say, up to 0.150 inch, are suitable for spinning monofils. In spinning from solution, the solution may be extruded either into an atmosphere heated up to or above the boiling point of the solvent or into a nonsolvent for the polymer, e.g., hexane.

After spinning, the polylactide polymer and copolymer filaments are drawn to effect orientation and to improve tensile strength. This is accomplished by drawing (permanently elongating) the filaments at a temperature between 50° C. and 140° C., preferably between 90° C. and 135° C. the preferred draw ratio being from 3:1 to 11:1. The drawing step may be conducted in one or more steps, in air or in a bath containing a liquid nonsolvent for the polymer, e.g., glycerol or water. This drawing brings about a marked increase in tensile strength and molecular orientation, as measured by the X-ray orientation angle.

Following the drawing, the filaments may be subjected to annealing. This may be carried out by running the oriented filaments from a feed roll to a takeup roll and heating the filaments between the rolls, with the takeup roll rotating at a speed ranging from the same speed of the feed roll to a speed 4 percent slower than that of the feed roll. At the first-mentioned speed ratio, essentially no shrinkage will take place, and at the second-mentioned speed ratio shrinkage will take place up to 4 percent of its length. As a consequence of this annealing, the filaments undergo essentially no shrinkage under the action of body fluids, when used as sutures.

Instead of spinning the polylactide polymers into filaments, it is possible to extrude or cast it into films, which are then drawn and annealed. The films thus treated can be cut into narrow strips for use as sutures. In the preferred embodiment the sutures are made from filaments.

As best illustrated in FIG. 1, if the polylactide filaments 12 are to be used for suturing, one end thereof may be inserted in a drilled needle 13 and securely fastened in place by swaging to form a needle and suture combination.

Polylactide filaments, unlike catgut, are adversely affected by moisture and tubing fluid. For this reason, polylactide prostheses are packaged dry in a hermetically sealed package a preferred form of which is shown in FIG. 2. Referring now to FIG. 2, there is shown a surgical package indicated generally as 14 having disposed therein a coil of polylactide suture 12 one end of which is attached to a needle 13. The needle and suture are positioned within a cavity 16 that is evacuated or

filled with a dry atmosphere such as nitrogen. The package is fabricated of two sheets of aluminum foil or aluminum foil plastic laminate material and heat sealed or bonded with adhesive at the skirt 18 to hermetically seal the cavity and isolate the contents of the package from the external atmosphere.

It is to be understood that minor amounts of inert additives such as coloring materials and plasticizers can be incorporated in the sutures by being mixed with the copolymers by known techniques. Any of a variety of plasticizers such as, for instance, glyceryl triacetate, ethyl benzoate, and diethyl phthalate can be used to advantage, especially with poly-L lactide. Preferred plasticizers for the glycolide copolymers are dibutyl phthalate and bis 2-methoxyethyl phthalate. The amount of plasticizer may vary from 1–40 percent based on the weight of the polymer. Not only does the plasticizer render the filaments more pliable and more easy to handle, but it also helps in spinning. By the term "inert" is meant materials that are inert chemically to the polymer, and are inert to living tissue, i.e., do not cause any of the adverse effects discussed on page 2 of this specification.

The present invention may be further illustrated by the following examples:

EXAMPLE I

Filaments having a diameter of 11.5 to 12.5 mils, a modulus of 1.04×10^6, tensile strength of 47,000 lb./sq. in., a knot strength of 37,000 lb./sq. in., and an elongation at break of 21 percent, were prepared by spinning polymer from L(−) lactide, said polymer having an inherent viscosity of 2.44 (measured at 0.1 percent concentration in benzene at 25° C.), from melt at 190° C., and drawing to 6:1 ratio in glycerol at 95° C. Some of the filaments were annealed taut at 126° C. and others at 100° C. as shown in more detail in table III which follows:

TABLE III

Annealed Taut at 126° C. for 5 Min	Shrinkage
Placed relaxed in oven at 126° C. for 5 minutes	7.4 percent
Control (i.e., not annealed)	28.2 percent
Placed relaxed in water at 100° C. for 5 minutes	13.0 percent
Control	28.2 percent
Placed relaxed in water at 77° C. for 5 minutes	1.4 percent
Control	18.0 percent
Annealed Taut at 100° C. for 5 Min	
Placed relaxed in oven at 100° C. for 5 minutes	11.0 percent
Control	21.4 percent
Placed relaxed in water at 77° C. for 5 minutes	7.4 percent
Control	18.0 percent

Annealed filaments such as described above are particularly useful as sutures as evidenced from example II.

EXAMPLE II

A polymer of L(−) lactide, said polymer having an inherent viscosity of 1.4, was melt spun at 160° to 170° C. into a monofilament. The filament was then drawn to four times the undrawn length by passage over a metal plate heated to 90° C. The filament obtained measured 0.007 inch in diameter. To improve dimensional stability, the drawn monofilament was annealed for 3 minutes at 90°–95° C. while under tension. The drawn, annealed filament was cut to convenient length and sterilized by being placed in polyethylene bags, which were sealed and exposed to two passes under a Van de Graaff beam of 2 million electron volts (1 to 1.5 Mrads per pass). Some of the bags contained dry monofilament, some contained monofilament in water and some contained monofilaments in ethyl alcohol.

The effect of annealing can be seen by these observations. When the annealed monofilament was heated in a dry oven at 95° C. for 3 minutes in a relaxed state, it shrank less than 4 percent. By contrast, an identical monofilament that had not been annealed shrank 25 percent. The annealed monofilament at 77° C. in water for 5 minutes shrank 14 percent.

In another experiment, the annealed monofilament was implanted in the abdominal cavity of a young adult male rat. After 16 days the implantation was removed. It had undergone less than 2 percent shrinkage.

The monofilaments thus obtained were used to connect severed muscle tissue in rats and in dogs in accordance with the following procedure:

A midline incision was made in the rat's abdominal skin, the skin was peeled back, and two small slits were then made in the abdominal muscles, one on either side of the midline. Each rat was sutured with several loops of the sterilized monofils prepared as above in one incision. Each rat had, as a control, either plain or chromic catgut suture in the other incision (size 4–0, 0.006–0.008 inch diameter). The skin was then closed and clamped. The rats were observed at regular intervals.

The sterilized monofils were tested for suturing dogs as follows: a midline incision about 3 to 4 inches long was made in the skin over the abdomen of a 6-month old dog. The skin was separated from the abdominal musculature and retracted with conventional retractors. Three incisions about 1 inch long were made through the abdominal musculature. One incision was closed with polylactic acid suture material, the other incisions were sutured with commercial catgut sutures (U.S.P. type A plain, size 4–0, and type C medium chromic, size 4–0).

Rats were sacrificed at intervals of 2, 4, 7, 14, 28, 59, 91, and 140 days. Dogs were sacrificed at 14, 23, and 50 days. In these examinations it was observed that the polylactic acid monofilaments were more slowly absorbed than plain gut. Further, it was observed that there was less general tissue reaction with the polylactic acid, as shown by gross appearance and by examination of histological sections.

EXAMPLE III

Polymer from L(−) lactide having an inherent viscosity of 3.11 at 0.1 percent concentration in benzene at 25° C., prepared by previously described methods, was converted to sutures by melt spinning, drawing, and annealing as described in example I.

EXAMPLE IV

Monofils of poly-DL-lactic acid, having an inherent viscosity of 1.42 at 0.1 percent concentration in benzene at 25° C., were tested as sutures after having been sterilized by two passages under a 2 Mev. electron beam at 1 to 1.25 Mrads. per pass. The sterilized monofils (0.006–0.008 inch diameter) were tested in suturing rats as follows:

A midline incision was made in the rat's abdominal skin, the skin was peeled back, and two small slits were then made in the abdominal muscles, one on either side of the midlines. Each rat was sutured with several loops of the sterilized monofils prepared as above in one incision and with a catgut suture as control in the other incision (unchromed, size 4–0, 0.006–0.008 inch diameter). The skin was then closed and clamped. The rats were observed at regular intervals. After approximately one month, the poly-DL-lactic acid sutures were about 50 percent hydrolyzed; tissue reaction was minimal to absent with no evidence of granuloma formation and adhesions. In the rats sutured with catgut, the catgut absorbed to about 60 percent after about 1 month, but there was pronounced tissue reaction with evidence of adhesions and granulation.

After about 60 days, both the polylactic acid and catgut sutures were absorbed, but the rats sutured with the catgut showed more scar tissue than the rats sutured with the polylactic acid.

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260-785

UNITED STATES PATENT OFFICE
CERTIFICATE OF CORRECTION

Patent No. 3,636,956 Dated January 25, 1972

Inventor(s) Allan K. Schneider

It is certified that error appears in the above-identified patent and that said Letters Patent are hereby corrected as shown below:

Col. 11, Table VI, under "Mole %", second occurrence, line 2 : -75- has been omitted.

Col. 11, Table VI, under "Grams", second occureence, line 2 : should read -71.4-.

Col. 12, line 8 : the degree sign should appear after "85".

Col. 12, line 59 : "Example 59" should read -Example XVIII-.

Col. 13, Table IX : the last number in the first line "Days post Implantation" should read -15-; the last number in second line should read -o.4-; the last number in last line should read -2.4-.

Col. 15, Claim 7 : "107" should read -1-.

Col. 15, Claim 15 : "114" should read -8-.

Signed and sealed this 28th day of November 1972.

(SEAL)
Attest:

EDWARD M.FLETCHER,JR.
Attesting Officer

ROBERT GOTTSCHALK
Commissioner of Patents

With rabbits, the suture material was found to be completely absorbed before about 38 days, irrespective of whether it was plain catgut or polylactic acid. However, the rabbits which were stitched with the polylactic acid sutures showed no adverse tissue reactions, with no tissue walling off or covering over of the suture material, in contrast to the behavior of catgut.

A midline incision of about 3 to 4 inches long was made in the skin over the abdomen of a 6-month old dog. The skin was separated from the abdominal musculature and retracted with conventional retractors. Two incisions about 1 inch long were made through the abdominal musculature. The right side incision was closed with polylactide suture material (size 4–0). The left incision was sutured with catgut (U.S.P. type A plain, size 4–0). After 4 days the polylactide was intact with no evidence of granulation or adhesion. At the end of 14 days, the dog was again examined and at the time the incision closed with the catgut showed intense inflammatory reaction. In contrast, the incision closed with the polylactic acid suture was free of granuloma formation, and the scar was clearly visible, i.e., no inflammation was evident. In both cases, however, the suture material had been absorbed by the tissue.

EXAMPLE V

A mixture of 95 parts by weight of L(−) lactide and 5 parts by weight of DL-lactide was fused under nitrogen, and there was added 0.125 part by weight of diethylzinc as a 25 percent solution in heptane. The mixture was heated at 105° C. for 1 hour at atmospheric pressure in an atmosphere of nitrogen. The solid L(−) lactide/DL-lactide (95/5) copolymer thus obtained had an inherent viscosity of 2.63 (0.1 percent solution in benzene at 34.5° C.). The copolymer was ground to a fine powder, which was in turn pressed to a plug suitable for use in an extrusion-spinning apparatus. Filaments of the copolymer were spun at about 200° C. through a 35 mil spinneret and were drawn to 6.4 times their original length in glycerol at about 120° C. The drawn filaments had the following properties:

Inherent viscosity	1.7
Diameter	12.5 mils
Tensile strength	58,500 p.s.i.
Elongation at break	20 percent
Modulus	1.08×10^6 p.s.i.
Knot strength	37,000 p.s.i.
Shrinkage after 5 minutes in water at 37° C	23 percent
Weight loss after 30 hours in boiling water	30 percent

After 90 Days in Distilled Water at 37° C

Inherent viscosity	0.53
Tensile strength	19,000 p.s.i.
Weight loss	2.8 percent

After 90 Days in Distilled Water at 37° C

Inherent viscosity	0.38
Tensile strength	9,000 p.s.i.
Weight loss	7 percent

EXAMPLES VI–XIII

A number of other L(−) lactide copolymers were prepared and spun into filaments by the method of example V. When the comonomer was a liquid at ordinary temperature (beta-propiolactone, gamma-butyrolactone, or pivalolactone), it was added to the lactide only after the lactide had been fused. The bulk polymer properties, spinning conditions, and filament properties of these copolymers are summarized in the following table:

TABLE IV

	Examples							
	VI	VII	VIII	IX	X	XI	XII	XIII
Weight percent comonomer	7.5% DL-lactide	10% DL-lactide	15% DL-lactide	5% glycolide	10% glycolide	5% beta-propiolactone	5% gamma-butyrolactone	5% pivalolactone
Inh. visc. (bulk)	2.87	2.50	2.30	2.53	2.52	1.31	2.99	2.68
Spinning temp., °C.	190	200	200	210	190	.....	170	170
Draw ratio	8.6	8.1	7.5	6.8	8.0	.....	10	10
Drawing temp., °C.	128	125	100	125	100	.....	115	100
Inh. visc. (drawn fil.)	1.75	1.75	1.47	1.84	1.70	.....	1.73	1.61
Diameter (mils)	11.5	11.0	12.5	10.3	8.5	.....	9.4	8.5
Ten. strength (p.s.i.)	53,300	69,000	53,000	77,000	57,800	.....	59,000	58,000
Elong. at break, percent	20	20.7	15.5	14	32	.....	17	22
Modulus (p.s.i.)	1.4×10^6	1.1×10^6	1.1×10^6	1.1×10^6	0.46×10^6	.....	.....	1.2×10^6
Knot strength (p.s.i.)	37,800	37,100	30,000	43,000	37,000	.....	42,000	48,000
Shrinkage (H₂O/77° C./5 min.), percent	13	37.5–44	7.2	12	7.3	.....	11.5	11.5
Wt. loss (H₂O/100° C./30 hrs.), percent	64	46	65	45	.....	.....	.....	.....
After 30 days in water at 37° C.:								
Inh. visc. (drawn fil.)		0.54	.....	0.56	.....	.....	0.91	0.72
Ten. strength (p.s.i.)		29,600	.....	46,000	.....	.....	.....	.....
Wt. loss, percent		3.5	.....	7.4	.....	.....	5.6	9.3
After 90 days in water at 37° C.:								
Inh. visc. (drawn fil.)				0.34	.....	.....	0.54	0.35
Ten. strength (p.s.i.)				12,000	.....	.....	.....	.....
Wt. loss, percent				12.1	.....	.....	7.1	3.5

[1] About.

EXAMPLES XIV–XV

Copolymers of L-lactide with the intermolecular cyclic esters of alpha-hydroxybutyric acid and alpha-hydroxyheptanoic acid were made by essentially the method of example V.

A mixture of 44.2 parts of L-lactide and 5.8 parts of the cyclic ester of alpha-hydroxybutyric acid was fused under nitrogen, and there was added 0.08 part of 25 percent solution of diethylzinc in heptane. The mixture was heated at 105°–108° C. for 3 hours at atmospheric pressure in an atmosphere of nitrogen. The resulting copolymer of L-lactide and the intermolecular cyclic ester of alpha-hydroxybutyric acid (88.4/11.6) had an inherent viscosity of 2.15 (0.1 percent solution in benzene).

The copolymer of L-lactide and the intermolecular cyclic ester of alpha-hydroxyheptanoic acid (90/10) was prepared similarly from 45 parts of L-lactide, 5 parts of cyclic ester, and 0.08 part of 25 percent solution of diethylzinc in heptane. After the mixture was heated for 3 hours, the resulting polymer had an inherent viscosity of 2.28.

The spinning conditions and filament properties of these copolymers are summarized in table V.

The intermolecular cyclic esters of alpha-hydroxybutyric acid and alpha-hydroxyheptanoic acid were prepared essentially by the method of Bischoff and Walden, Ann. 279, 100 (1895). The sodium salts of the corresponding alpha-bromo acids were made from the acids and sodium methoxide in an ethyl ether/ethyl alcohol mixture. The cyclic esters were made by heating the sodium salts to 300°–315° C. under reduced pressure. The butyric acid derivative was purified by distillation at 78°–85° C./0.07 mm. and by crystallization from ethyl alcohol/petroleum ether, with cooling in solid carbon dioxide. The heptanoic acid derivative was purified by crystallization from pentane, with cooling in solid carbon dioxide, and from

ethyl alcohol. Both cyclic esters were characterized by elemental analyses and infrared absorption spectra

TABLE V

Example	XIV	XV
% Comonomer (by weight)	11.6% intermolecular cyclic ester of alpha-hydroxybutyric acid	10% intermolecular cyclic ester of alpha-hydroxyheptanoic acid
Inh. viscosity	2.15	2.28
(bulk)		
Spinning temperature	185° C	190° C
Draw ratio	10*	8
Drawing temperature	94° C., 122° C.*	98° C
Inherent viscosity	1.62**	1.63**
Tensile strength (p.s.i.)	66,300	59,100
Elongation at break	22.3%	18.3%
Modulus (p.s.i.)	1.04×10^5	0.95×10^5
Shrinkage (H_2O/77° C./5 min.)	20.6%	55.0%
Weight loss (H_2O/100° C./48 hrs.)	60.5%	63.6%

*The filament was drawn in two stages. In the first stage, it was drawn 7X (draw ratio) at 94° C., in the second, it was drawn at 122° C. to an extent sufficient to give an overall draw ratio of 10.

**Measured on undrawn filament. The inherent viscosities of the filaments of examples I - XIII were measured on drawn filaments.

EXAMPLE XVI

A mixture of 206 g. of powdered L-lactide/DL-lactide (90/10) copolymer and 0.6182 g. of the monosodium salt of 4-{4-(N-ethyl-p-sulfobenzylamino)diphenylmethylene}-{1-(N-ethyl-N-p-sulfoniumbenzyl)-$\Delta^{2,5}$-cyclohexadienimine}[F D & C (Food, Drug, and Cosmetic) Green No. 1] was rotated in a Fisher-Kendall mixer for 48 hours at room temperature. The resulting homogeneous mixture was pressed to a plug and spun into green monofilaments by essentially the method of example V.

EXAMPLE XVII

The weighed amounts of L(−) lactide melting at 98°–99° C. and having a specific rotation (sodium D-line, 25° C.) of −295 to −300, and glycolide (m.p. 82.8–84.5° C.) are mixed in the quantities indicated below and added to a cylindrical tube containing stannous octoate catalyst and a magnetic stir bar. After sealing under 110 mm. of mercury pressure, the vessel is heated at 105° C. for 96 hours with magnetic stirring to yield a cylinder of solid copolymer. In each case, 0.0039 mol (0.1580 g.) of stannous octoate is used as the catalyst. The monomer initiator ratio (A/I) is 1,500.

The reacting quantities and mol percent of the comonomers investigated in this example are summarized in the following table:

TABLE VI

GLYCOLIDE Mole %	GLYCOLIDE Grams	GLYCOLIDE Moles	LACTIDE Mole %	LACTIDE Grams	LACTIDE Moles
20	13.3	0.116	80	44.6	0.46
25*	19.1	0.163		77.8	0.69
30	20.2	0.17	70	58.6	0.41
35	23.6	0.20	65	54.7	0.38
40	26.7	0.23	60	50.4	0.35
45	30.2	0.26	55	46.1	0.32
50	33.4	0.29	50	41.8	0.29
55	39.2	0.34	45	40.1	0.28
60	47.1	0.41	30	25.1	0.17

A similar series is run using tetraphenyl tin as a catalyst at an A/I of 2,000 with similar results.

Each copolymer (from 20 mole percent glycolide to 70 mole percent glycolide) is extruded under pressure at a temperature of 10°–220° C. through a 35-mil orifice. The extruded fiber has a diameter of 33–36 mils and is drawn to five times its original length. The extruded fibers are heated to 70°–85° C. during this drawing step.

Strong resilient fibers having excellent tensile and dry knot strength are thus obtained, the physical characteristics of these fibers being summarized in the following table

*Used 0.1780 g. of catalyst

TABLE VII

Glycolide, mole percent	Diameter, mils	Straight, lbs.	Diameter, mils	Dry knot, lbs.	Initiator	K/S
20	15.9	8.87	15.4	6.98	Ph	0.82
20	12.4	6.00	13.1	6.82	O	0.66
25	14.7	9.14	15.2	6.61	O	0.73
30	14.8	8.66	15.2	6.17	Ph	0.66
30	13.1	9.28	13.1	6.80	O	0.80
30	14.1	9.28	14.8	6.77	O	0.84
34	14.6	10.2	14.8	6.34	Ph	0.76
40	14.3	8.38	14.2	6.63	O	0.64
40	14.2	10.2	14.3	6.7	O	0.80
46	14.1	11.4	14.1	7.12	Ph	0.73
50	14.8	10.0	14.7	6.78	O	0.84
60	14.1	11.2	13.8	6.58	Ph	0.76
60	14.2	8.26	14.0	4.58	Ph	0.56

NOTE—Ph=tetraphenyl tin, monomer/initiator ratio=2,000. O=stannous octoate, monomer/initiator ratio=1,000. K/S=dry knot straight pull ratio.

The biological behavior of the L(−)-lactide glycolide copolymers prepared in accordance with the present example is summarized in table VIII. Sections of suture material are implanted subcutaneously in rats and removed at various intervals to determine changes in tensile strength and diameter. A large increase in the diameter of a suture following implantation is an indication of shrinkage (dimensional instability).

TABLE VIII

Glycolide, mole percent		Days post implantation 0	1	8	10	15
20	Tensile strength (lbs.)	10.1	8.8	6.7	4.2	4.0
	Diameter (mils)	14.8	15.4	16.6	16.4	16.4
25	Tensile strength (lbs.)	10.1	8.8	6.0	3.3	3.0
	Diameter (mils)	14.0	14.7	15.4	15.6	16.9
30	Tensile strength (lbs.)	10.3	8.2	6.4	3.0	1.6
	Diameter (mils)	14.1	15.4	22.7	23.8	24.2
35	Tensile strength (lbs.)	10.1	6.0	2.6	.....	.....
	Diameter (mils)	12.4	20.0	24.9	.....	.....
40	Tensile strength (lbs.)	8.6	3.4	2.2	0.0	.....
	Diameter (mils)	14.0	22.8	27.4	28.1	.....
50	Tensile strength (lbs.)	8.9	3.2	1.4	0.0	.....
	Diameter (mils)	15.0	24.6	29.4	41.0	.....
60	Tensile strength (lbs.)	8.6	2.8	0.2	.....	.....
	Diameter (mils)	14.1	20.2	26.4	.....	.....

EXAMPLE VIII

Fifty-four and seven-tenths parts by weight of L(−) lactide (0.38 mols) melting at 98°–99° C. and having a specific rotation (sodium D-line, 25° C.) of −295 to −300 is mixed with 23.6 parts by weight (0.20 mols) of glycolide (m.p. 82.8°–84.5° C.) and 0.0039 mol (0.158 parts by weight) of stannous octoate in a dry PYREX glass flask containing a stir bar under dry nitrogen. The monomer initiator ratio (A/I) is 1,500. The glass flask is sealed under 110 mm. of mercury pressure and the vessel is heated at 105° C. for 96 hours with magnetic stirring to yield a solid copolymer.

The 35 mole percent glycolide-lactide copolymer so obtained is extruded under pressure at an elevated temperature through a 35 mil orifice and drawn to five times the original length. The extruded fiber is heated to 70°–85° C. during this drawing step. The biological behavior of this 35 mole percent copolymer in rats is summarized in table IX

TABLE IX

Days post implantation	0	1	6	10	1
Tensile strength, lbs	10.0	8.9	4.2	3.1	0
Tensile strength, p.s.i. X10³	60.0	53.3	25.2	18.6	2

A 35 mole percent copolymer, prepared as described above may be extruded to form a rod that can be oriented by drawing 3× at an elevated temperature. The rod so formed will have a tensile strength greater than 25,000 p.s.i.

Although this invention has been specifically illustrated with monofilaments, the products of the present invention may also be manufactured in the form of multifilaments, that may be braided to form sutures. Filaments suitable for braiding having a diameter in the range of 0.00025–0.003 inches may be conveniently obtained by dry spinning a L(−) lactide polymer dissolved in a suitable solvent. The manufacture of a braided size 2/0 suture from multifilament obtained by dry spinning a L(−) lactide copolymer is illustrated in the following example.

EXAMPLE XIX

A round-bottomed PYREX flask having a long neck is carefully cleaned, flame dried, evacuated and purged two times with dry nitrogen. To the flask is added under a dry nitrogen atmosphere:

231.42 parts glycolide (M.P. 82.8–84.5° C.) 30.19 wt. percent
533.52 parts L(−) lactide (M.P. 98°–99° C.) 69.61 wt. percent
1.5558 parts stannous octoate, 0.20 wt. percent

The flask is evacuated to 125 mm. pressure and heated at 105° C. for 66 hours. The polymer so obtained (inherent viscosity in 0.1 percent chloroform solution = 3.2–3.4) is dissolved in dry 1, 1,2-trichloroethane (distilled from phosphorous pentoxide) to give a clear 8 percent (W/W) solution (bulk viscosity 1,600 poise).

The spin dope (8 percent solution) is heated to 90° C. and extruded through a 10-hole 0.005 inch spinneret (capillary land/diameter = 2.4) at a rate of 3 milliliters per minute into a heated column 15 feet long and 6 inches in diameter. The temperature within the heated column varied from 128° C. at the bottom to 142° C. at the top and the column is swept with hot nitrogen (131–134° C.) at a rate of 5 cubic feet per minute. The extruded filaments are taken up on a reel at a linear speed of 150 feet per minute. The inherent viscosity of the filamentary material is 3.4 indicating no degradation during the spinning process. The copolymer filament is lustrous in appearance and has the following physical characteristics:

Tensile Strength	1.0 grams/denier
Elongation	330 percent
Young's Modulus	24 grams/denier

The filament contains about 1.5 percent residual solvent.

The yarn from the takeup spool is 6-plied to 60 filaments and drawn 4.5 times at 75° C. and 25 feet per minute input speed through a tubular furnace swept with nitrogen. The drawn yarns prepared as described above have the following physical characteristics:

Tensile Strength	2.8–3.3 grams/denier
Elongation	26 percent
Young's Modulus	50 grams/denier

The individual filaments have a tensile strength of about 4.5–5.0 grams/denier; an elongation of about 38 percent and a Young's Modulus of about 45 grams per denier. The yarn is braided to form a size 2/0 braided suture, packaged in a dry atmosphere in a hermetically sealed container and sterilized by Cobalt 60 gamma irradiation. The in vivo absorption characteristics of this braided suture material in rats are indicated in Table X.

TABLE X

	After days post implantation				
	0	1	6	10	15
Tensile strength (X10³ p.s.i.)	88	47	37	31	30

It will be noted from a comparison of table IX and table X that the braided structure obtained from yarn that has been dry spun from a suitable solvent (example XIX) retained in vivo tensile strength for a longer period of time than a melt extruded monofilament of similar composition (example XVIII). The improvement in in vivo tensile strength exhibited by the dry spun braided suture is such that the amount of glycolide in the lactide copolymer composition may be increased to 40 mole percent (34.8 weight percent glycolide, 65.2 weight percent L-lactide). A copolymer suture of this composition (40 mole percent glycolide 60 mole percent L-lactide) has tensile strength and absorption characteristics similar to catgut.

Multifilament yarns that contain polylactide filaments together with nonabsorbable filaments of DACRON, TEFLON, nylon, etc., are useful in the manufacture of vascular grafts. Such a multifilament yarn is illustrated in FIG. 4 wherein the nonabsorbable fiber is represented by the hatched fiber cross section 19. In FIG. 4, the fibers 20 are extruded from lactide polymer and copolymer compositions as described above. The relative proportions of absorbable filaments 20 and nonabsorbable filaments 19 may be varied to obtain the absorption characteristic desired in the woven fabric or tubular implants. Methods of weaving and crimping vascular prostheses are described in U.S. Pat. No. 3,096,560.

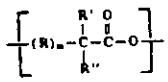
Composite fabrics of absorbable and nonabsorbable materials fashioned by textile processes including weaving, knitting, and fabricating by the nonwoven felting of fibers are described in U.S. Pat. No. 3,108,357 and U.S. Pat. No. 3,463,158. Similar techniques may be used in the manufacture of surgical aids wherein nonabsorbable fibers are combined with absorbable fibers composed of lactide polymers and copolymers. The surgical utility of "bicomponent filaments" containing absorbable and nonabsorbable components is described in U.S. Pat. No. 3,463,158, the teaching of which is incorporated herein by reference. Monofilaments of lactide polymers and copolymers may be woven or knitted to form an absorbable fabric having the structure illustrated in FIG. 5, useful surgically in hernia repair and in supporting damaged liver, kidney, and other internal organs.

The products of the invention are useful in surgical applications where an absorbable aid or support is required, for example, in the formation of surgical mesh, absorbable staple, artificial tendons, or cartilage material, and in other uses where a temporary aid during healing is needed. They may also be used to advantage in repairing hernias and in anchoring organs which have become loose.

As many apparently widely different embodiments of this invention may be made without departing from the spirit and scope thereof, it is to be understood that this invention is not limited to the specific embodiments thereof except as defined in the appended claims.

I claim:

1. A sterile surgical suture absorbable without causing unfavorable tissue reaction and essentially dimensionally stable within the body comprising an oriented synthetic polylactide polymer containing more than about 85 percent by weight of repeating units of one antipodal species of alpha-hydroxypropionic acid and no more than about 15 percent by weight of repeating units of the formula



where R is lower alkylene, m is an integer of 0 to 1, R' is selected from the class consisting of hydrogen and lower alkyl, and R'', which can be the same or different than R', is selected from the class consisting of hydrogen and alkyl of up to 22 car-

bons when m is 0 and, when m is 1, R'' is selected from the class consisting of hydrogen and lower alkyl, said polylactide before being oriented being characterized by having an inherent viscosity of at least 1.2 at 0.1 percent concentration in benzene at 25° C. and by losing at least about 20 percent of its weight on treatment with boiling water for a period of 100 hours, and being further characterized by exhibiting a tensile strength of from 40,000 to about 100,000 p.s.i. and by having a diameter of 0.0005–0.045 inches.

2. The suture of claim 1, packaged in a dry atmosphere within a hermetically sealed container.

3. The suture of claim 1, packaged within an evacuated hermetically sealed container.

4. The suture of claim 1, wherein the polylactide polymer is a poly L(−) lactide containing up to 15 percent by weight of repeating units derived from DL-lactide.

5. The suture of claim 1, containing a minor amount of inert coloring agent and plasticizer.

6. The suture of claim 1, containing bis 2-methoxyethyl phthalate as a plasticizer.

7. A method of retaining living tissue in a desired location and relationship during a healing process which comprises:

sewing living tissue with the suture of claim 107, whereby said suture becomes imbedded in the tissue;

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process.

8. The suture of claim 1, having a sterile needle attached to one end thereof.

9. The needle and suture combination of claim 8, packaged in a dry atmosphere within a hermetically sealed container.

10. The needle and suture combination of claim 8, packaged within an evacuated hermetically sealed container.

11. The needle and suture combination of claim 8, wherein said monofilament is a poly-L-(−) lactide containing up to 15 percent by weight of repeating units derived from DL-lactide.

12. The needle and suture combination of claim 8, wherein said monofilament is a 95/5 weight percent copolymer of L(−) lactide and DL-lactide.

13. The needle and suture combination of claim 8, wherein said monofilament contains a minor amount of inert coloring agent and plasticizer.

14. The needle and suture combination of claim 8, wherein said monofilament contains bis 2-methoxyethyl phthalate as a plasticizer.

15. A method of retaining living tissue in a desired location and relationship during a healing process which comprises:

sewing living tissue with the suture of claim 114, whereby said suture becomes imbedded in the tissue;

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process.

16. A sterile surgical suture absorbable without causing unfavorable tissue reaction and essentially dimensionally stable within the body comprising a synthetic polylactide copolymer containing at least about 65 mole percent of repeating units derived from one antipodal species of alpha-hydroxypropionic acid and not more than about 35 mole percent of repeating units derived from alpha-hydroxyacetic acid, said polylactide being characterized by having an inherent viscosity of at least 1.2 at 0.1 percent concentration in a suitable solvent at 25° C. and by losing at least about 20 percent of its weight on treatment with boiling water for a period of 100 hours, and being further characterized by exhibiting a tensile strength of from 40,000 p.s.i. to about 100,000 p.s.i. and by having a diameter of 0.0005–0.045 inches.

17. The suture of claim 16, packaged in a dry atmosphere within a hermetically sealed container.

18. The suture of claim 16, packaged within an evacuated hermetically sealed container.

19. The suture of claim 16, wherein the polylactide copolymer contains about 35 mole percent of repeating units derived from alpha-hydroxyacetic acid.

20. The suture of claim 16, wherein the polylactide copolymer contains about 30 mole percent of repeating units derived from alpha-hydroxyacetic acid.

21. The suture of claim 16, wherein the polylactide copolymer contains about 25 mole percent of repeating units derived from alpha-hydroxyacetic acid.

22. The suture of claim 16, wherein the polylactide copolymer contains about 20 mole percent of repeating units derived from alpha-hydroxyacetic acid.

23. The suture of claim 16, wherein the polylactide copolymer contains about 15 weight percent of repeating units derived from alpha-hydroxyacetic acid.

24. The suture of claim 16, wherein the polylactide copolymer contains about 5 weight percent of repeating units derived from alpha-hydroxyacetic acid.

25. The suture of claim 16, containing a minor amount of inert coloring agent and plasticizer.

26. The suture of claim 16, containing bis 2-methoxyethyl phthalate as a plasticizer.

27. A method of retaining living tissue in a desired location and relationship during a healing process which comprises:

sewing living tissue with the suture of claim 16, whereby said suture becomes imbedded in the tissue;

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process.

28. The suture of claim 16, having a sterile needle attached to one end thereof.

29. The needle and suture combination of claim 28, packaged in a dry atmosphere within a hermetically sealed container.

30. The needle and suture combination of claim 28, packaged within an evacuated hermetically sealed container.

31. The needle and suture combination of claim 28, wherein the polylactide polymer contains about 35 mole percent of repeating units derived from alpha-hydroxyacetic acid.

32. The needle and suture combination of claim 28, wherein the polyactide polymer contains about 30 mole percent of repeating units derived from alpha-hydroxyacetic acid.

33. The needle and suture combination of claim 28, wherein the polylactide polymer contains about 25 mole percent of repeating units derived from alpha-hydroxyacetic acid.

34. The needle and suture combination of claim 28, wherein the polylactide polymer contains about 20 mole percent of repeating units derived from alpha hydroxyacetic acid.

35. The needle and suture combination of claim 28, wherein the polylactide polymer contains about 15 weight percent of repeating units derived from alpha-hydroxyacetic acid.

36. The needle and suture combination of claim 28, wherein the polylactide polymer contains about 5 weight percent of repeating units derived from alpha-hydroxyacetic acid.

37. The needle and suture combination of claim 28, wherein said monofilament contains a minor amount of inert coloring agent and plasticizer.

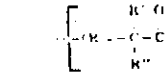
38. The needle and suture combination of claim 28, wherein said monofilament contains bis 2-methoxyethyl phthalate as a plasticizer.

39. A method of retaining living tissue in a desired location and relationship during a healing process which comprises:

sewing living tissue with the needle and suture combination of claim 28, whereby said suture becomes imbedded in the tissue;

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process.

40. A sterile surgical suture absorbable without causing unfavorable tissue reaction and essentially dimensionally stable within the body in the form of a braided structure, comprising filaments of a synthetic polylactide polymer containing at least about 85 percent by weight of repeating units of one antipodal species of alpha-hydroxypropionic acid and no more than about 15 percent by weight of repeating units of the formula



where R is lower alkylene, m is an integer of 0 to 1, R' is selected from the class consisting of hydrogen and lower alkyl, and R'', which can be the same or different from R', is selected from the class consisting of hydrogen and alkyl group of up to 22 carbons when m is 0 and, when m is 1, R'' is selected from the class consisting of hydrogen and lower alkyl, said polylactide being characterized by having an inherent viscosity of at least 1.2 at 0.1 percent concentration in benzene at 25° C. and by losing at least about 20 percent of its weight on treatment with boiling water for a period of 100 hours, at least 50 percent of the filaments making up the braided structure being oriented and the diameter of the filaments ranging from 0.00025 to 0.003 inches; and the braided structure itself being characterized by exhibiting a tensile strength of from 40,000 p.s.i. to about 100,000 p.s.i.

41. The suture of claim 40, packaged in a dry atmosphere within a hermetically sealed container.

42. The suture of claim 40, packaged within an evacuated hermetically sealed container.

43. The suture of claim 40, wherein the polylactide copolymer filaments that make up said braided structure contain up to 15 percent by weight of repeating units derived from DL-lactide.

44. The suture of claim 40, wherein the polylactide copolymer filaments that make up said braided structure are a 95/5 weight percent copolymer of L(−) lactide and DL-lactide.

45. The suture of claim 40, wherein the filaments that make up said braided structure contain a minor amount of inert coloring agent and plasticizer.

46. The suture of claim 40, wherein the filaments that make up said braided structure contain bis 2-methoxyethyl phthalate as a plasticizer.

47. A method of retaining living tissue in a desired location and relationship during a healing process which comprises:

sewing living tissue with the suture of claim 40, whereby said suture becomes imbedded in the tissue;

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process.

48. The suture of claim 40, having a sterile needle attached to one end thereof.

49. The needle and suture combination of claim 48, packaged in a dry atmosphere within a hermetically sealed container.

50. The needle and suture combination of claim 48, packaged within an evacuated hermetically sealed container.

51. The needle and suture combination of claim 48, wherein the polylactide copolymer filaments that make up said braided structure contain up to 15 percent by weight of repeating units derived from DL-lactide.

52. The needle and suture combination of claim 48, wherein the polylactide copolymer filaments that make up said braided suture are a 95.5 weight percent copolymer of L(−) lactide and DL-lactide.

53. The needle and suture combination of claim 48, wherein the filaments that make up said braided suture contain a minor amount of inert coloring agent and plasticizer.

54. The needle and suture combination of claim 48, wherein the filaments that make up said braided suture contain bis 2-methoxyethyl phthalate as a plasticizer.

55. A method of retaining living tissue in a desired location and relationship during a healing process which comprises

sewing living tissue with the suture and needle combination of claim 47, whereby said suture becomes imbedded in the tissue,

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process

56. A sterile surgical suture absorbable without causing unfavorable tissue reaction and essentially dimensionally stable within the body in the form of a braided structure comprising filaments of a synthetic polylactide copolymer containing at least 60 mole percent of repeating units derived from one antipodal species of alpha-hydroxypropionic acid and no more than 40 mole percent of repeating units derived from alpha-hydroxyacetic acid, said polylactide being characterized by having an inherent viscosity of at least 1.2 at 0.1 percent concentration in a suitable solvent at 25° C. and by losing at least about 20 percent of its weight on treatment with boiling water for a period of 100 hours, at least 50 percent of the filaments making up the braided structure being oriented, and the braided structure itself being further characterized by exhibiting a tensile strength of at least 15,000 p.s.i. 10 days following implantation in an animal body

57. The suture of claim 56, packaged in a dry atmosphere within a hermetically sealed container

58. The suture of claim 56, packaged within an evacuated hermetically sealed container

59. The suture of claim 56, wherein the polylactide copolymer contains about 35 mole percent of repeating units derived from alpha-hydroxyacetic acid

60. The suture of claim 56, wherein the polylactide copolymer contains about 30 mole percent of repeating units derived from alpha-hydroxyacetic acid.

61. The suture of claim 56, wherein the polylactide copolymer contains about 25 mole percent of repeating units derived from alpha-hydroxyacetic acid.

62. The suture of claim 56, wherein the polylactide copolymer contains about 20 mole percent of repeating units derived from alpha-hydroxyacetic acid

63. The suture of claim 56, wherein the polylactide copolymer contains about 15 weight percent of repeating units derived from alpha-hydroxyacetic acid

64. The suture of claim 56, wherein the polylactide copolymer contains about 5 weight percent of repeating units derived from alpha-hydroxyacetic acid.

65. The suture of claim 56, wherein the filaments that make up said braided structure contain a minor amount of inert coloring agent and plasticizer.

66. The suture of claim 56, wherein the filaments that make up said braided structure contain bis 2-methoxyethyl phthalate as a plasticizer.

67. A method of retaining living tissue in a desired location and relationship during a healing process which comprises:

sewing living tissue with the suture of claim 56, whereby said suture becomes imbedded in the tissue;

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process.

68. The suture of claim 56, having a sterile needle attached to one end thereof.

69. The needle and suture combination of claim 68, packaged in a dry atmosphere within a hermetically sealed container.

70. The needle and suture combination of claim 68, packaged within an evacuated hermetically sealed container.

71. The needle and suture combination of claim 68, wherein the polylactide copolymer filaments that make up said braided structure contain about 35 mole percent of repeating units derived from alpha-hydroxyacetic acid.

72. The needle and suture combination of claim 68, wherein the polylactide copolymer filaments that make up said braided structure contain about 30 mole percent of repeating units derived from alpha-hydroxyacetic acid.

73. The needle and suture combination of claim 68, wherein the polylactide copolymer filaments that make up said braided structure contain about 25 mole percent of repeating units derived from alpha-hydroxyacetic acid

74. The suture of claim 68, wherein the polylactide copolymer filaments that make up said braided structure contain about 20 mole percent of repeating units derived from alpha-hydroxyacetic acid

75. The needle and suture combination of claim 68, wherein the polylactide copolymer filaments that make up said braided structure contain about 15 weight percent of repeating units derived from alpha-hydroxyacetic acid

76. The needle and suture combination of claim 68, wherein the polylactide copolymer filaments that make up said braided

structure contain about 5 weight percent of repeating units derived from alpha-hydroxyacetic acid

77. The needle and suture combination of claim 68, wherein the filaments that make up said braided structure contain a minor amount of inert coloring agent and plasticizer

78. The needle and suture combination of claim 68, wherein the filaments that make up said braided structure contain bis 2-methoxyethyl phthalate as a plasticizer

79. A method of retaining living tissue in a desired location and relationship during a healing process which comprises

sewing living tissue with the needle and suture combination of claim 68, whereby said suture becomes imbedded in the tissue,

and leaving the suture in said tissue until said suture is absorbed by the tissue during the healing process

* * * * *

United States Patent [19]

Shalaby et al.

[11] 4,141,087

[45] Feb. 27, 1979

ISOMORPHIC COPOLYOXALATES AND SUTURES THEREOF

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Assignee: Ethicon, Inc., Somerville, N.J.

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U.S. Cl. ..... 3/1; 3/1.4; 3/1.5; 128/92 B; 128/92 C; 128/334 R; 128/335.5; 260/860; 528/307

Field of Search ..... 260/75 R, 860; 3/1, 3/1.4, 1.5; 128/92 B, 92 C, 334 R, 335.5

References Cited

U.S. PATENT DOCUMENTS

11,251	2/1937	Carothers	260/78 A
11,762	3/1938	Ellis	260/75 R
12,652	9/1960	Beindorff et al.	260/30.6
13,901	5/1975	Coquard et al.	128/335.5
12,993	7/1977	Coquard et al.	128/335.5

OTHER PUBLICATIONS

Korshak, V.V., et al., Acad. Sci. USSR Bull., Division of Chemical Sciences, pp. 889–893 (1957).

Korshak, V.V., et al., "Polyesters", Pergamon Press, N.Y., N.Y., pp. 31–63 (1965).

Primary Examiner—Benjamin R. Padgett

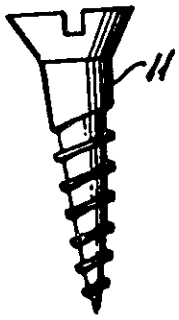
Assistant Examiner—T. S. Gron

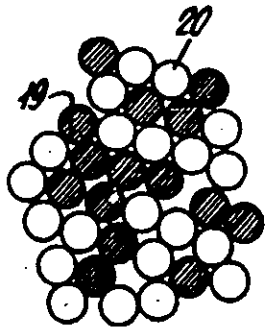
Attorney, Agent, or Firm—Wayne R. Eberhardt

[57] ABSTRACT

Synthetic absorbable sutures are prepared from copolyoxalate polymers having isomorphic sequences. The polymers are derived from mixtures of cyclic and linear diols, each having the same carbon chain length of 6 or 8 atoms. The cyclic diol may be aliphatic or aromatic. The diols are polymerized with dialkyl oxalate, preferably in the presence of an inorganic or organometallic catalyst, to obtain a highly crystalline isomorphic copolyoxalate polymer which is melt extruded and drawn to form oriented filaments. The filaments are characterized by good initial tensile and knot strength and a high order of softness and flexibility. When implanted in living animal tissue, the fibers have good strength retention over a period of at least 21 days and eventually absorb with a minimal degree of adverse tissue reaction.

21 Claims, 5 Drawing Figures





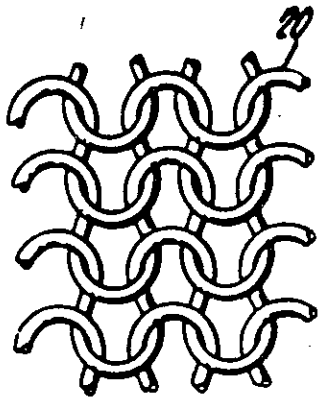


Fig. 1.

Fig. 2.

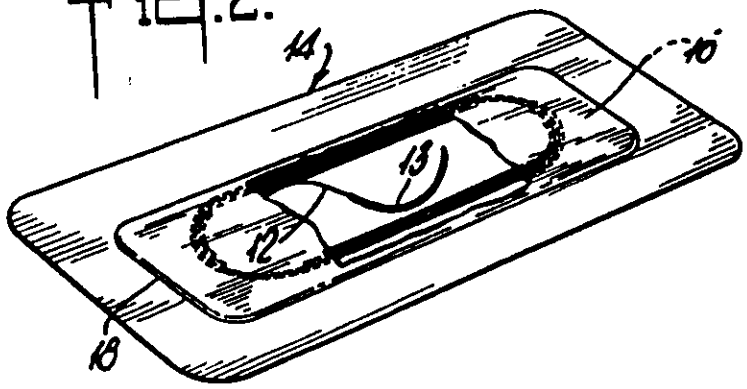


Fig. 3.

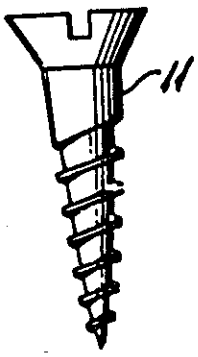


Fig. 4.

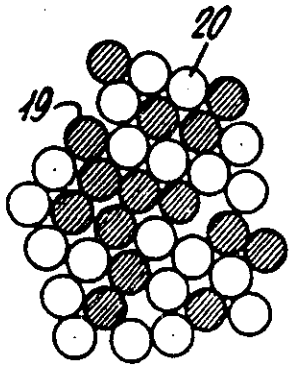
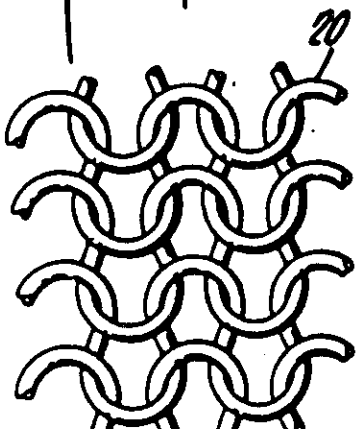


Fig. 5.



ISOMORPHIC COPOLYOXALATES AND SUTURES THEREOF

BACKGROUND OF THE INVENTION

1. Field of the Invention

This invention relates to synthetic absorbable sutures, and more particularly, to synthetic absorbable sutures comprising extruded and oriented filaments of copolymers of polyoxalates having isomorphic sequences.

2. Description of Prior Art

Absorbable suture materials have traditionally been natural collagenous materials obtained from sheep or beef intestine, commonly known as catgut. More recently, it has been proposed to manufacture synthetic absorbable sutures from polyesters of hydroxycarboxylic acids, notably polylactide, polyglycolide, and copolymers of lactide and glycolide. Such synthetic absorbable sutures are described in U.S. Pat. Nos. 3,636,956 3,297,033 and elsewhere in the literature. Polyesters of succinic acid have also been suggested for at least partially bioresorbable surgical articles as disclosed for example in U.S. Pat. No. 3,883,901.

Among the requirements of an ideal absorbable suture are that it should have good handling properties, should approximate and hold tissue for proper healing with minimal tearing and tissue damage, should have adequate straight tensile and knot strength, should be controllably uniform in properties including dimensional stability within the body, should be sterilizable, should be absorbable by living tissue, preferably at a constant rate regardless of the place in the body or the condition of the patient and without causing such unfavorable tissue reactions as walling off, granuloma formation or excessive edema, and finally should be capable of being properly and easily tied into surgical knots.

While multifilament sutures manufactured from polymers of lactide and glycolide fulfill the above requirements to a large degree, monofilament sutures of these materials are considerably less flexible than catgut and these synthetic sutures are accordingly generally limited to a multifilament, braided construction. Sutures of glycolide polymers are also not suitable for sterilization by radiation without suffering severe degradation of physical properties.

We have discovered that copolyoxalate copolymers having isomorphic sequences can be melt extruded into pliable, monofilament fibers which have good in vivo strength retention and are absorbed in animal tissue without significant adverse tissue reaction. The fibers have good tensile and knot strength, and can be sterilized by gamma radiation without serious loss of these properties. In addition, monofilament sutures of the polymers of the present invention have a high degree of softness and flexibility not found in many synthetic absorbable sutures of the prior art.

The preparation of polyoxalate polymers is described in the art. Carothers et al, J. Amer. Chem. Soc. 52, 3292 (1930) for example, describes the ester interchange reaction of diols such as ethylene glycol, 1,3-propanediol, or 1,4-butanediol with diethyl oxalate to yield a mixture of monomer, soluble polymer and insoluble polymer. The reaction of oxalic acid and an alkylene glycol to form polyester resins is described in U.S. Pat. No. 2,111,762, while the preparation of polyesters of fiber-forming quality from dicarboxylic acids and diols is described in U.S. Pat. Nos. 2,071,250-1 and 2,952,652. Isomorphic polymers including polyester copolymers have been discussed in the literature[1]. The particular isomorphic copolyoxalates of the present invention however, have not previously been known, nor has their preparation or use as synthetic absorbable sutures been suggested heretofore.

It is accordingly an object of the present invention to provide new and useful polymers of isomorphic copolyoxalates and articles made therefrom. A further object of this invention is to provide synthetic absorbable sutures of isomorphic copolyoxalates. It is a yet further object of this invention to provide surgical aids and prostheses fabricated of fibers or cast or machined from blocks of isomorphic copolyoxalate polymers.

SUMMARY

Highly crystalline isomorphic polyoxalate polymers are prepared by reacting mixtures of cyclic and linear diols with dialkyl oxalate, preferably in the presence of an inorganic or organometallic catalyst. The diols comprising the reaction mixture have the same carbon chain length separation between terminal OH groups of 6 or 8 carbon atoms. The cyclic diol may be trans 1,4-cyclohexane dialkanol or p-phenylene dialkanol and comprises

(1) Isomorphism in Synthetic Macromolecular Systems, G. Allegra and L. W. Bassi, Adv. Polymer Sci. 6, 549 (1969) from about 5 to 95 mol percent, and preferably from 40 to 75 r ol percent of the total diol reactant.

Copolymers prepared by the transesterification reaction of the two diols and diethyl oxalate are melt extruded into highly crystalline filaments suitable for use as synthetic absorbable sutures. Drawn and oriented filaments are characterized by high tensile and knot strength, a Young's modulus in most cases of less than about 600,000 psi providing a high order of filament softness and flexibility, and good strength retention and minimal tissue reaction in vivo.

DESCRIPTION OF DRAWINGS

FIG. 1 is a perspective view of a needle-suture combination;

FIG. 2 is a perspective view of a needle-suture combination within a hermetically sealed container;

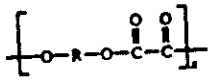
FIG. 3 illustrates a screw machined from the polymer of the present invention;

FIG. 4 is a cross-sectional view of a composite yarn containing filaments of different composition and;

FIG. 5 is a plan view of a surgical fabric knitted from fibers of the present invention.

DESCRIPTION OF PREFERRED EMBODIMENTS

Polymers of the present invention are comprised of isomorphic units of cyclic and linear oxalates and have the general formula



wherein each R is

$$-(CH_2)_a-A-(CH_2)_a- \quad \text{I}$$

or

$-(CH_2)-_{a+2a}$ II

with from about 5 to 95 mol percent, and preferably from about 40 to 75 mol percent of R groups being I; A is trans 1,4-cyclohexylene or p-phenylene, a is 1 or 2 and is the same for I and II, and x is the degree of polymerization resulting in a fiber forming polymer having a molecular weight greater than about 10,000.

Polymers of the present invention are conveniently prepared by an ester interchange reaction between the afore-described mixture of diols and a lower ester of oxalic acid, preferably in the presence of an ester interchange catalyst. The preferred ester of oxalic acid is diethyl oxalate. The ester interchange is most preferably conducted in two stages wherein the reactants are first heated with stirring under a nitrogen atmosphere to form a prepolymer with the removal of ethanol, followed by postpolymerization under heat and reduced pressure to obtain a final polymer of the desired molecular weight and fiber forming quality. Polymers with low or moderate degrees of polymerization are postpolymerized in the liquid state or as finely-divided solid particles, depending on their melting temperature range.

The polymer is melt extruded through a spinnerette in a conventional manner to form one or more filaments which are subsequently drawn about 4X to 6X in order to achieve molecular orientation and improve tensile properties. The resulting oriented filaments have good tensile and dry knot strength and good in vivo strength retention.

It is well documented that the crystallinity and hence suitability for fiber-formation in both the AB and AA-BB type polyesters decreases significantly when the mol fraction of the major comonomer sequence decreases below about 80%. In some instances, if the comonomer sequences are isomorphic, chains composed of slightly less than 80% of the major sequences can pack into a crystalline form. However, randomly constructed copolyester chains based on almost equal amounts of the isomorphic comonomer sequences are generally found to be non-crystalline and poor fiber formers. Contrary to this general rule, the isomorphic copolyesters of the present invention display an unexpectedly high level of crystallinity of about 45% in a 50/50 copolyester. The polymers of the present invention are also unusual in that all copolymers through the entire composition range of from 5 to 95% of each isomorphic comonomer demonstrate levels of crystallinity comparable to those encountered in the parent homopolymers; namely between 30 and 50% depending on the thermal history. A similarly striking observation characteristic of these copolyesters is their display of melting endotherms, as shown by DSC, for the crystalline regions of all copolymers within the composition range of from about 5 and 95 mol % of each isomorphic comonomer. Constructed curves of the melting temperature versus composition did not reveal any positive eutectic composition in these systems. The X-ray and DSC data suggest strongly the uncommon presence of almost complete isomorphism in the copolyesters of the present invention.

Dimensional stability and tensile strength retention of the oriented filaments may be enhanced by subjecting the filaments to an annealing treatment. This optional treatment consists of heating the drawn filaments to a temperature of from about 40° to 130° C., most preferably from about 60° to 110° C. while restraining the filaments to prevent any substantial shrinkage. The filaments are held at the annealing temperature for a few seconds to several days or longer depending on the temperature and processing conditions. In general, annealing at 60° to 110° C. for up to about 24 hours is satisfactory for the polymers of the present invention. Optimum annealing time and temperature for maximum fiber in vivo strength retention and dimensional stability is readily determined by simple experimentation for each fiber composition.

Filaments of the present invention may be used as sutures in either a monofilament or a multifilament construction. Multifilament sutures are preferably braided but may also be twisted or covered in accordance with common practice. For use as sutures, it is necessary that the fibers be sterile, and sterilization may be accomplished by exposing the fibers to Cobalt 60 gamma radiation or to ethylene oxide. Such sterilization techniques are well known and commonly practiced in suture manufacture.

Since the function of a suture is to join and hold severed tissue until healing is well along, and to prevent wound separation as a result of movement or exercise, a suture must meet certain minimum standards of strength. It is particularly important that strength be maintained when knots are tied and during the actual procedure of drawing tight a suitable knot. Sutures prepared from oriented filaments of the present invention are characterized by a straight tensile strength of at least about 30,000 psi and a knot strength of at least about 20,000 psi, although significantly higher strengths may be obtained.

The preparation of high molecular weight oriented filaments of isomorphic polyoxalates is further illustrated by the following examples where all percentages are on a molar basis unless otherwise noted. The following analytical methods were used to obtain the data reported in the examples. Inherent viscosity (η_{inh}) was obtained on polymer solutions (1 gram/liter) in chloroform or hexafluoro-2-propanol (HFIP). The infrared spectra of polymer films (cast from $CHCl_3$ or HFIP) were recorded on a Beckman Acculab 1 spectrophotometer. The NMR spectra of the polymer solutions in $CHCl_3$ were recorded on an MH-100 or CFT-20 spectrophotometer. A DuPont 990 DSC apparatus was used to record the glass transition (T_g), crystallization (T_c) and melting (T_m) temperatures of the polymers under nitrogen, using about 5 mg samples and a heating rate of 10° C./min. or as otherwise specified. The thermogravimetric analysis (TGA) data of the polymers were recorded under nitrogen using a DuPont 950 TGA apparatus and a heating rate of 10° or 20° C./min. with about 10 mg samples. A Philips vertical goniometer with graphite crystal monochromatized copper K_α radiation was used to obtain the X-ray powder and fiber diffraction patterns of the polymers. Crystallinity was determined by the method of Hermans and Weidinger and the diffractometer patterns were resolved with a DuPont 310 curve analyzer.

In vitro hydrolysis of polymer discs (about 1.2 g, 2.2 cm diameter) and monofilaments (7–25 mil) was conducted at 37° C. in phosphate buffer comprising a solution of 27.6 g sodium dihydrogenphosphate monohydrate in 1000 ml. water adjusted to pH 7.25 with sodium hydroxide.

In vivo absorption (muscle) was determined by implanting two 2 cm segments of monofilament fiber into the left gluteal muscles of female Long Evans rats. The implant sites were recovered after periods of 60, 90, 120

180 days and examined microscopically to deter- the extent of absorption. In vivo absorption (sub- eous) is a non-histological technique in which nuous observation of the biological degradation of ents of suture is made by implanting two segments iture, 2 cm long, into the abdominal subcutis of g female rats. The implants are readily visible n the skin is wetted with propylene glycol and it of absorption can be determined by subjective l examination.

vivo strength retention was determined by im- ting segments of sutures in the posterior dorsal utis of female Long Evans rats for period of 5 to 30 . The sutures were recovered at the designated ds and pull-tested for straight tensile strength.

vitro strength retention was determined by placing ents of sutures in the afore-defined buffer at 50° C. periods of 2 to 4 days. The sutures were recovered he designated periods and pull-tested for straight ile strength.

EXAMPLES

General Polymerization Procedure

iethyl oxalate was heated with selected diols in a hanically-stirred reactor using a stannous alkanoate organic titanate catalyst. The reaction was con- ted under a nitrogen atmosphere at suitable temper- es until a substantial portion of the calculated ount of ethanol was obtained. Postpolymerization of resulting prepolymer was then continued under uced pressure using a suitable heating scheme. At end of the postpolymerization period, the molten ymer was allowed to cool slowly at room tempera- , isolated, ground and dried at 25° C. to 80° C. (de- ding on the polymer T_m) in vacuo for at least one . Alternatively, the prepolymer can be postpolymer- d partially in the liquid state, cooled, and then post- lymerized further in the solid state as finely divided ticles. Detailed experimental conditions for the prep- ation of representative samples of isomorphic polyox- tes and important properties of the resulting poly- rs are presented below.

EXAMPLE I

95/5 Poly (trans 1,4-Cyclohexylenedicarbinyl-co-hexamethylene Oxalate)

Distilled diethyl oxalate (19.0 g, 0.130 mol), recrystal- ed trans 1,4-cyclohexanedimethanol (19.8 g, 0.137 ol), 1,6-hexadiol (0.856 g, 0.00724 mol) and stannous toate (0.33 M in toluene; 0.080 ml, 0.026 mmol) were ded under dry and oxygen-free conditions to a glass actor equipped for magnetic stirring. The prepolymer as formed by heating the mixture at 120° C. for 3 ours under nitrogen at 1 atmosphere while allowing e formed ethanol to distill, followed by heating at 60° C. for 2 hours. The prepolymer was then heated in acuo (0.05 mm Hg) at 220° C. for 1 hour, and the ostpolymerization completed by heating at 215° C. for n additional 6 hours. The polymer was then allowed to ool to room temperature, isolated and ground, and inally dried in vacuo at room temperature.

Polymer Characterization:

ηinh in $CHCl_3$ = 0.50

DSC (20° C./min.): T_m = 210° C.

Polymer Melt-Spinning:

The polymer was spun using an Instron Rheometer with a 30 mil die at 207° C.

In Vitro Evaluation:

The undrawn fibers lost 21 and 66 percent of their initial mass after immersion in phosphate buffer at 37° C. for 42 and 127 days, respectively.

EXAMPLE II

85/15 Poly (1,4-Cyclohexylenedicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (58.4 g, 0.400 mols), recrystallized trans 1,4-cyclohexanedimethanol (less than 1% cis isomer; 53.9 g, 0.374 mols), 1,6-hexanediol (7.8 g, 0.066 mol), and stannous octoate (0.33M in toluene; 0.40 ml, 0.13 mmol) were added under dry and oxygen-free conditions to a glass reactor equipped for mechanical stirring. The mixture was heated at 120° and 150° C. for 2 and 3 hours, respectively, under nitrogen at one atmosphere while the formed ethanol distilled. The prepolymer was allowed to cool, then reheated to 200° C. under reduced pressure (0.1 mm Hg). Temperatures of 200°, 220° and 230° C. were maintained for 2, 3 and 4 hours while the collection of distillates continued. The resulting polymer (ηinh in $CHCl_3$ = 0.49) was cooled, isolated, ground (2 mm screen size), and then dried in vacuo at room temperature. Portions (30 g) of this ground polymer were postpolymerized in the solid state in glass reactors equipped for magnetic stirring by heating in vacuo (0.1 mm Hg) at 185° C. for 22 hours.

Polymer-Characterization:

ηinh in $CHCl_3$ = 1.14

DSC (20° C./min.): T_m = 187° C.

Polymer Melt-Spinning:

The polymer was spun at 230° C. using an Instron Rheometer with a 40 mil die. The fiber was quenched in ice water, wound, dried and subsequently drawn.

Fiber Properties:

Fibers drawn 5X in two stages, 4X at 62° C. followed by 1.25X at 119° C. exhibited the following properties: diameter = 8.5 mils, straight tensile strength = 8.39×10^4 psi; knot tensile strength = 5.06×10^4 psi; modulus = 6.61×10^5 psi; elongation = 15%.

In Vivo Evaluation:

Sterilized (via γ-radiation, 2.5 Mrads), drawn monofilament (8.5 mils) retained 89, 75, 10 and zero percent of its initial breaking strength (4.8 lbs.) after subcutaneous implantation in rat muscle for 3, 7, 14 and 21 days respectively. Drawn filaments implanted into the gluteal muscles of rats elicited median tissue responses in the slight range throughout a 180 day post-implantation period. Filaments drawn 4X at 60° C. followed by 1.25X at 110° C. and having a straight tensile of 6.76×10^4 psi showed indications of initial degradation 20 to 26 weeks after implantation.

In Vitro Evaluation:

Fibers drawn 4X at 60° C. (exhibiting a straight tensile of 4.33×10^4 psi) lost 40 percent of their initial mass after immersion in phosphate buffer at 37° C. for 84 days.

EXAMPLE III

80/20 Poly (1,4-Cyclohexylenedicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (43.8 g, 0.300 mol), recrystallized trans 1,4-cyclohexanedimethanol (cis isomer con-

tent = 1.0%, 36.3 g, 0.252 mol), 1,6-hexanediol (7.4 g, 0.063 mol), and stannous oxalate (12.4 mg., 0.060 mmol) were added under dry and oxygen-free conditions to a glass reactor equipped for mechanical stirring. The prepolymer was formed by heating the mixture at 120° C. for 2 hours under nitrogen at 1 atmosphere while allowing the formed ethanol to distill, followed by 160° C. for 2.5 hours. The mixture was allowed to cool, then reheated in vacuo (0.1 mm Hg) to 140° C. and maintained until the prepolymer melted. The temperature was then increased to 190° C., maintained for 30 minutes, then raised to 200° C. for 1.5 hours. The melt post-polymerization of the stirred polymer was completed by heating at 220° C. for 4.5 hours. The polymer was cooled, isolated, ground (screen size = 2 mm) and dried in vacuo at room temperature. To obtain the final product, the ground polymer was post-polymerized in the solid state in a glass reactor equipped for magnetic stirring by heating at 180° C. in vacuo (0.05 mm Hg) for 24 hours while allowing the formed diols to distill.

Polymer Characterization:

ηinh in $CHCl_3$ = 1.33

DSC (20° C./min.): T_m = 205° C.

Polymer Melt-Spinning:

The polymer was spun at 240° C. using an Instron Rheometer equipped with a 40 mil die. The extruded filaments were quenched in ice water, wound, then dried at room temperature in vacuo, and subsequently drawn 4X.

Fiber Properties:

Diameter = 9.0 mils; straight tensile strength = 7.31×10^4 psi; knot tensile strength = 3.46×10^4 psi; modulus = 7.7×10^5 psi; elongation = 15%.

In Vivo Evaluation:

Sterilized (by γ-radiation, 2.5 Mrads), fibers (9.0 mil) retained 85, 20 and zero percent of their initial breaking strength (4.2 lbs.) after subcutaneous implantation in rat muscles for 3, 7 and 14 days, respectively. These fibers were also implanted into the gluteal muscles of rats to determine tissue response and absorption characteristics. The median tissue response elicited by the samples was in the slight range after 5 days post implantation and in the minimal range after 42 days; absorption of the samples was first noted at 120 days and by 180 days approximately fifty percent of the material had been absorbed.

EXAMPLE IV

80/20 Poly (1,4-Cyclohexylenedicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (23.4 g, 0.160 mol), recrystallized trans 1,4-cyclohexanedimethanol (cis isomer content = 6.3%; 20.0 g, 0.139 mol), 1,6-hexandiol (4.1 g, 0.035 mol) and Tyzor OG* (0.117M in toluene, 0.28 ml, 0.033 mmols) were added under dry and oxygen-free conditions to a glass reactor equipped for magnetic stirring. A prepolymer was formed by heating the mixture at 120° C. for 19 hours under nitrogen at 1 atmosphere while allowing the formed ethanol to distill. The pressure was then reduced (0.05 mm Hg) and heating at 120° C. continued for 30 minutes longer. The temperature was then increased and maintained at 180° C., 190° C. and 200° C. for 2, 3 and 2 hours, respectively, while removing excess and formed diols. The polymer was allowed to cool, isolated, ground, and dried in vacuo at room temperature.

*Tyzor OG, a titanium glycolate catalyst manufactured by E. I. DuPont de Nemours and Co., Wilmington, Del., 19898

Polymer Characterization:

ηinh in $CHCl_3$ = 0.46

DSC (10° C./min.): T_m = 171° C.

TGA (10° C./min. under N_2): 0.25% weight lost at 275° C.

Polymer Melt-Spinning:

The polymer was spun using an Instron Rheometer with a 30 mil die at 172° C. The extruded filaments were quenched in ice water, dried in vacuo at room temperature, and finally drawn 5X at 43° C.

Fiber Properties:

ηinh in $CHCl_3$ = 0.42

X-ray: Major reflections correspond to 8.9 (W), 4.84 (M), 4.41 (S) and 3.42Å (W) d-spacings; 26% crystallinity. (Undrawn filaments were found to be 22% crystalline which increased to 31% by annealing at 70° C. for one hour).

Physical Properties:

Diameter = 11.1 mils; straight tensile strength = 2.07×10^4 psi; elongation = 35%.

In Vivo Evaluation:

The rate of absorption and tissue response of drawn filaments was determined by implantation into the ventral abdominal subcutis of Long-Evans rats. Some evidence of filament degradation was noted 11 to 14 weeks after implantation, with the bulk of the fiber being absorbed by 20 to 23 weeks. No tissue reaction to the implants was noted at any period.

In Vitro Evaluation:

The drawn fibers exhibited a 43% decrease in mass after immersion in the phosphate buffer at 37° C. for 28 days.

EXAMPLE V

67/33 Poly(trans 1,4-cyclohexylenedicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (40.0 g, 0.274 mol), recrystallized trans 1,4-cyclohexanedimethanol (25.9 g, 0.180 mol), 1,6-hexanediol (10.6 g, 0.0897 mol), and stannous octoate (0.33 M in toluene; 0.16 ml. 0.053 mmol) was added to a glass reactor equipped for mechanical stirring. The prepolymer was formed by heating the mixture under nitrogen at 120° C. for 9 hours, followed by 125° C. for 9 hours while collecting the distillates. The prepolymer was cooled, then reheated in vacuo (0.03 mm Hg) and maintained at 80, 120, 150, 170 and 180° C. for 1, 2, 2, 3 and 1.5 hours, respectively. The postpolymerization of the polymer melt was completed by heating at 195° C. for 6 hours while continuing to stir and remove distillates. The polymer was cooled, isolated, ground, and then dried at room temperature.

Polymer Characterization:

ηinh in $CHCl_3$ = 0.49

DSC (20° C./min.): T_m = 179° C.

Polymer Melt Spinning:

The polymer was spun at 175° C. using an Instron Rheometer with a 30 mil die. The resulting fibers were subsequently drawn 4X at 50° C.

Fiber Properties:

Diameter = 9.3 mils, straight tensile strength = 2.65×10^4 psi, knot tensile strength = 2.21×10^4 psi, modulus = 3.7×10^5 psi.

In Vivo Evaluation, Tissue Reaction:

Two centimeter long samples of sterilized (by γ-radiation, 2.5 Mrads) drawn fiber were implanted sub-

cutaneously in the abdominal wall of young female Long Evans strain rats. At intervals of 3, 14, 28, 56 and 90 days, two rats were sacrificed for examination of implants. The skin containing the fibers was excised and affixed to plastic sheets for preservation in formalin. Two tissue blocks were cut transversely from each site and embedded in paraffin for histologic preparation. Eight stained samples were examined at each interval for tissue reaction to the fibers. Only mild foreign body reactions were detected.

In Vivo Evaluation, Absorption:

Fiber segments sterilized by γ-radiation (2.5 Mrads) approximately 2 cm in length were inserted into the ventral abdominal subcutis of Long Evans rats (100 g, female) to determine the rate of absorption of the drawn fibers. One to two rats were sacrificed after various periods after implantation. The skin containing the implant sites was removed and dried. These preparations were examined and evaluated using both dissecting and transmission microscopes. Estimates of the amount of implant remaining were based on the length of the segment or fragments remaining and the decrease in the surface area made by palpating the implant in the dried hide and comparing it with a one week old preparation. Implants were fragmented at one week; migration and clumping of fragments was noted at subsequent kill periods. Evidence of degradation was first seen 16 weeks after implantation. Palpable fragments, in diminishing amounts, were present until 30 weeks. Quantitatively, about 100, 75, 45, 40, 20, 15 and 5 or less percent of the suture remained after 14, 16, 20, 23, 26, 30 and 36 weeks.

EXAMPLE VI

50/50 Poly (trans 1,4-cyclohexyldicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (38.0 g, 0.260 mol), recrystallized trans 1,4-cyclohexanedimethanol (20.2 g, 0.140 mol), 1,6-hexanediol (16.5 g, 0.140 mol), and stannous octoate (0.33 M in toluene, 0.16 ml, 0.053 mmol) were added under dry and oxygen-free conditions to a mechanically stirred glass reactor. Under nitrogen at one atmosphere, the mixture was heated to and maintained at 120° C. for 20 hours, while allowing the formed ethanol to distill. The prepolymer was cooled and then reheated in vacuo (0.05 mm Hg) to and maintained at 80°, 120°, 140°, 165°, 175°, 185°, and 195° C. for 1, 1, 3, 3.5, 2, 1 and 1 hour respectively. The removal of the diols was continued by heating at 200° C. for 8 hours to complete the postpolymerization. The polymer was cooled, isolated, ground, and then dried in vacuo at room temperature.

Polymer Characterization:

η_{inh} in $CHCl_3$ = 0.36

DSC (20° C./min.): T_m = 138° C.

Polymer Melt Spinning:

The polymer was spun at 136° C. using an Instron Rheometer (40 mil die) and was immediately drawn 5X at 53° C.

Fiber Properties:

X-ray Data: Major reflections correspond to 8.9 (W), 4.84 (M), 4.41 (S), and 5.40 Å (W) d-spacings; 36% crystallinity.

Physical Properties: Diameter = 10.6 mils, straight tensile strength = 1.36×10^4 psi, knot tensile strength = 1.13×10^4 psi, modulus = 1.33×10^5 psi, elongation = 27%.

In Vivo Evaluation:

Sterilized (by γ-radiation) drawn fiber segments (2 centimeters in length) were implanted into the ventral abdominal subcutis for study of the rate of absorption and tissue reaction.

At one week the implants were fragmented, clumping, and migrating, with the bulk of the suture being absorbed between 6 to 11 weeks. Thereafter, fragments with scattered birefringent particles or birefringent particles in a shell-like outline were observed. The birefringent particles decreased in amount until at 36 weeks only a few widely scattered particles were noted.

Only mild foreign body reactions were observed to be elicited by the sterilized drawn fiber segments during the test intervals of 3, 14, 28, 48, 90 and 180 day post implantation.

In Vitro Evaluation:

Undrawn fibers exhibited a 57 percent decrease in their initial mass after immersion in phosphate buffer at 37° C. for 28 days.

EXAMPLE VII

50/50 Poly (trans 1,4-cyclohexyldicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (58.5 g, 0.400 mol), recrystallized trans 1,4-cyclohexanedimethanol (cis isomer content = 0.7%; 29.7 g, 0.206 mol), 1,6-hexanediol (24.3 g, 0.206 mol), and stannous oxalate (16.5 mg, 0.080 mmols), were added under dry and oxygen-free conditions to a mechanically stirred glass reactor. The mixture was heated under nitrogen at one atmosphere to and maintained at 120° and 160° C. for 3 and 2 hours respectively while allowing the formed ethanol to distill. The propolymer was cooled and then reheated in vacuo (0.05 mm Hg) and maintained at 170°, 190° and 205° C. for 3, 2.5 and 3 hours respectively while continuing to remove excess and formed diol to complete the postpolymerization. The polymer was cooled, isolated, ground, and then dried in vacuo at room temperature.

Polymer Characterization:

η_{inh} in HFIP = 1.07

DSC (20° C./min.) T_m = 132° C.

Polymer Melt Spinning:

The polymer was spun at 150° C. using an Instron Rheometer (40 mil die) and was drawn 4X at 50° C. followed by 1.5X at 72° C.

Fiber Properties:

X-ray Data: Major reflections correspond to 9.11 (MS), 4.82 (S), 4.60 (W), 4.37 (S) and 3.45 Å (W) d-spacings; 46% crystallinity.

Physical Properties: Diameter = 7.6 straight-tensile strength = 51,300 psi, knot tensile strength = 36,400 psi, elongation = 31%.

EXAMPLE VIII

30/70 Poly (trans 1,4-cyclohexylenedicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (36.5 g, 0.250 mol), recrystallized trans 1,4-cyclohexanedimethanol (11.5 g, 0.0797 mol), 1,6 hexanediol (22.4 g, 0.190 mol), and stannous octoate (0.33 M in toluene; 0.16 ml, 0.053 mmol) were added under dry and oxygen-free conditions to a mechanically stirred reactor. The mixture was heated to and maintained at 125°, 140° and 160° C. for 2, 2 and 1 hour, respectively, under nitrogen at one atmosphere

while allowing the formed ethanol to distill. The prepolymer was cooled and then reheated in vacuo (0.1 mm Hg) and maintained at 150° and 185° C. for 16 and 3 hours, respectively. The postpolymerization was completed by maintaining the polymer at 200° C. for 5.5 hours while continuing to remove the diols under vacuum. The polymer was then cooled, isolated, ground and dried in vacuo at room temperature.

Polymer Characterization:

ηinh in $CHCl_3$ = 0.82

DSC (20° C./min): T_m = 85° C.

Polymer Melt Spinning:

The polymer was spun at 125° C. using an Instron Rheometer with a 40 mil die. The fiber was quenched in ice water, wound, dried in vacuo at room temprature, and subsequently drawn 5.6X at room temperature, followed by annealing at 55° C.

Fiber Properties:

Diameter 8.3 mils, straight tensile strength 5.18×10^4 psi, knot tensile strength 3.51×10^4 psi, modulus 2.11×10^5 and elongation 50%.

In Vivo Evaluation:

Sterilized (by γ-radiation, 2.5 Mrads), drawn fibers (9.8 mil diameter; 3.64×10^4 psi straight tensile strength; 2.34×10^4 psi knot tensile strength; 1.47×10^5 psi modulus; and an elongation of 45%) were implanted into the gluteal muscles of rats to determine their absorption and tissue response characteristics at 5, 21, 42 and 150 days post implantation.

At the 42 day period, there was no evidence of any morphologic changes of the implant sites indicating absorption. At the 150 day period, the fibers had a median value of 2 percent suture cross sectional area remaining (with a range of 0 to 20 percent).

Foreign body tissue responses to the samples were in the slight range at 5, 21 and 42 day periods and in the minimal range at the 150 day period.

In Vitro Evaluation:

Drawn fibers possessing physical properties similar to those of fibers used in the in vivo testing exhibited a 100% decrease in their initial mass after 141 days of immersion in phosphate buffer at 37° C.

EXAMPLE IX

5/95 Poly (trans 1,4-cyclohexylenedicarbinyl-co-hexamethylene Oxalate):

Distilled diethyl oxalate (19.0 g, 0.130 mol), recrystallized trans 1,4-cyclohexanedimethanol (1.0 g, 0.0069 mol), 1,6-hexanediol (16.3 g, 0.138 mol), and stannous octoate (0.33 M in toluene; 0.08 ml, 0.026 mmol) were added under dry and oxygen-free conditions to a glass reactor equipped for magnetic stirring. The prepolymer was formed by heating the mixture at 120° C. for 3 hours under nitrogen at one atmosphere while allowing the formed ethanol to distill, followed by 160° C. for 2 hours. The prepolymer was heated and maintained at 205° C. for 8 hours in vacuo (0.05 mm Hg). The polymer was then cooled, isolated, ground, and dried at room temperature.

Polymer Characterization:

ηinh in $CHCl_3$ = 0.88

DSC (20° C./min): T_m = 69° C.

TGA (20° C./min. under N_2): Less than 0.5% weight loss at 275° C. was recorded.

Polymer Melt Spinning:

The polymer was spun in an Instron Rheometer using 30 mil die at 85° C. The fibers were quenched in ice water and subsequently drawn 5X at room temperature.

Fiber Properties:

Diameter = 14.7 mils, straight tensile strength = 1.36×10^4 psi, knot tensile strength = 1.41×10^4 psi, modulus = 4.8×10^4 psi, elongation = 90%.

In Vitro Evaluation:

The drawn fibers exhibited a 93 percent decrease in their initial mass after immersion in phosphate buffer at 37° C. for 42 days.

EXAMPLE X

58/42 Poly (1,4-phenylenedicarbinyl-co-hexamethylene Oxalate):

Diethyl oxalate (14.6 g, 0.100 mols), recrystallized 1,4-benzenedimethanol (6.9 g, 0.050 mols), 1,6-hexanediol (8.3 g, 0.070 mols), and Tyzor TOT* catalyst (0.4 ml of a 1% solution) were added under dry and oxygen-free conditions to a glass reactor equipped for stirring. The prepolymer was formed by heating under nitrogen at one atmosphere at 140° C. for 4 hours while allowing the formed ethanol to distill. The mixture was then heated in vacuo (0.1 mm Hg) at 165° C. for 22 hours while continuing to remove distillates. A postpolymerization was conducted at 180°, 190°, and 200° C. for 2, 1 and 4 hours respectively. The polymer was cooled, ground and dried.

*Tyzor TOT, a tetraalkyl titanate catalyst manufactured by E. I. DuPont de Nemours and Co., Wilmington, Delaware, 19898.

Polymer Characterization:

ηinh in HFIP = 0.48

DSC (10° C./min): T_m = 170° C.

TGA (10° C./min in N_2): Less than 1% cummulative weight loss experienced at 250° C.

Polymer Melt Spinning:

The polymer was spun at 166° C. using an Instron Rheometer equipped with a 30 mil die.

In Vitro Evaluation:

Immersion of a molded disc, 2.2 cm in diameter, for 8 and 78 days in phosphate buffer at 37° C. resulted in a loss of 3 and 99 percent of the initial mass, respectively.

EXAMPLE XI

56/44 Poly (1,4-phenylenedicarbinyl-co-hexamethylene Oxalate):

Dibutyl oxalate (20.2 g, 0.100 mols), 1,4-benzenedimethanol (8.3 g, 0.060 mols), 1,6-hexanediol (5.6 g, 0.047 mols), and tetraisopropylorthotitanate catalyst (0.3 ml, of a 0.01M solution) were added under dry and oxygen-free conditions to a glass reactor equipped for magnetic stirring. The prepolymer was formed by heating at 140°, and 160° C. for 1, and 17 hours respectively under nitrogen at one atmosphere while allowing the formed butanol to distill. The pressure was reduced (0.2 mm Hg) while continuing to heat at 160° C. for an additional hour. The postpolymerization of the polymer melt was completed by heating at 180° C. and 200° C. for 2, and 3.5 hours, respectively, while continuing to remove distillates. The polymer was cooled, and isolated.

Polymer Characterization:

ηinh in HFIP = 0.42

DSC (10° C./min): T_m = 165° C.

TGA (10° C./min in N_2): Less than 1% cummulative weight loss experienced at 250° C.

In Vitro Evaluation:

Immersion of a molded disc, 2.2 cm in diameter, for 7 and 77 days, in phosphate buffer at 37° C. resulted in a loss of 3 and 56 percent of the initial mass, respectively.

EXAMPLE XII

50/50 Poly (1,4-phenylenedicarbinyl-co-hexamethylene Oxalate):

In a manner similar to that employed in Examples X and XI, the above identified copolymer having the following characteristics was produced:

DSC (10° C./min): T_m = 175° C.

TGA (10° C./min, in N_2): Less than 1% cummulative weight loss experienced at 250° C.

In Vitro Evaluation:

Immersion of a molded disc, 2.2 cm in diameter, for 8 and 78 days in phosphate buffer at 37° C. resulted in a loss of 6 and 54 percent of the initial mass, respectively.

While the preceding examples have been directed to the preparation of specific copolymers of polyoxalates, these examples are for purposes of illustration only and are not limiting of the invention. Mixtures of these polymers and combinations of these polymers with up to about 50 percent by weight of poly (alkylene oxalates) and other compatible polymers which produce nontoxic and absorbable polymers are likewise included within the present invention.

It is to be understood that inert additives such as coloring materials and plasticizers can be incorporated in the sutures. As used herein, the term "inert" means materials that are chemically inert to the polymer and biologically inert to living tissue, i.e., do not cause any of the adverse effects previously discussed. Any of a variety of plasticizers such as, for instance, glyceryl triacetate, ethyl benzoate, diethyl phthalate, dibutyl phthalate and bis-2-methoxyehtyl phthalate can be used if desired. The amount of plasticizer may vary from 1 to about 20 percent or more based on the weight of the polymer. Not only does the plasticizer render the filaments of the present invention even more pliable, it also serves as a processing aid in extrusion and thread preparation.

Filaments of the present invention are adversely affected by moisture and are accordingly preferably stored in hermetically sealed and substantially moisture-free packages, a preferred form of which is shown in FIG. 4. In FIG. 2, there is shown a suture package 14 having disposed therein a coil of suture 12, one end of which is attached to needle 13. The needle and suture are positioned within a cavity 16 that is evacuated or filled with a dry atmosphere of air or nitrogen. The illustrated package is fabricated of two sheets of aluminum foil or an aluminum foil-plastic laminate and heat sealed or bonded with adhesive at the skirt 16 to hermetically seal the cavity and isolate the contents of the package from the external atmosphere.

Filaments of the present invention may be used as monofilament or multifilament sutures, or may be woven, braided, or knitted either alone or in combination with other absorbable fibers such as poly (alkylene oxalate), polyglycolide or poly (lactide-co-glycolide), or with nonabsorbable fibers such as nylon, polypropylene, polyethylene-terephthalate, or polytetrafluoroethylene to form multifilament sutures and tubular structures having use in the surgical repair of arteries, veins, ducts, esophagi and the like.

Multifilament yarns that contain isomorphic copolyoxalate filaments of the present invention together with nonabsorbable filaments are illustrated in FIG. 4 wherein the nonabsorbable fiber is represented by the hatched fiber cross-section 19. In FIG. 4, the fibers 20 are extruded from polymer compositions of the present invention as described above. The relative proportions of absorbable filaments 20 and nonabsorbable filaments 19 may be varied to obtain the absorption characteristic desired in the woven fabric or tubular implants.

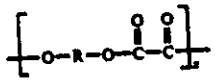
Composite fabrics of absorbable and nonabsorbable materials fashioned by textile processes including weaving, knitting and nonwoven felting are described in U.S. Pat. Nos. 3,108,357 and 3,463,158. Methods of weaving and crimping tubular vascular prostheses are described in U.S. Pat. No. 3,096,560. Similar techniques may be used in the manufacture of surgical aids wherein nonabsorbable fibers are combined with absorbable fibers composed of the polymers of this invention. The surgical utility of "bi-component filaments" containing absorbable and nonabsorbable components is described in U.S. Pat. No. 3,463,158 the teaching of which is incorporated herein by reference. Monofilaments of the polymers of the present invention may be woven or knitted to form an absorbable fabric having the structure illustrated in FIG. 5, useful surgically in hernia repair and in supporting damaged liver, kidney and other internal organs.

The polymers of the present invention are also useful in the manufacture of cast films and other solid surgical aids such as scleral buckling prostheses. Thus, cylindrical pins, screws as illustrated in FIG. 3, reinforcing plates, etc., may be machined from solid polymer having in vivo absorption characteristics depending upon the polymer composition and molecular weight.

Many different embodiments of this invention will be apparent to those skilled in the art and may be made without departing from the spirit and scope thereof. It is accordingly understood that this invention is not limited to the specific embodiments thereof except as defined in the appended claims.

We claim:

1. A synthetic absorbable suture of oriented fiber comprising an isomorphic polyoxalate polymer consisting essentially of units of cyclic and linear oxalates and having the general formula



wherein each R is

$$-(CH_2)_n - A - (CH_2)_n- \quad \text{I}$$

or

$$-(CH_2)_{n+1}- \quad \text{II}$$

and from about 5 to 95 mol percent of the R units are I; A is trans 1,4-cyclohexylene or p-phenylene, n is 1 or 2 and is the same for I and II, and x is the degree of polymerization resulting in a fiber forming polymer having a molecular weight greater than about 10,000.

2. A suture of claim 1 wherein said fiber is a monofilament.

3. A suture of claim 1 wherein said fiber is a multifilament.

4. A suture of claim 3 wherein said multifilament fiber is a braid.

5. A suture of claim 1 wherein n is 1 and A is trans 1,4-cyclohexylene.

6. A suture of claim 1 wherein n is 2 and A is trans 1,4-cyclohexylene.

7. A suture of claim 5 wherein from about 40 to 75 mol percent of the R units are of formula I.

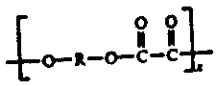
8. A suture of claim 1 wherein n is 1 and A is p-phenylene.

9. A suture of claim 1 wherein n is 2 and A is p-phenylene.

10. A suture of claim 1 having a surgical needle attached to at least one end thereof.

11. A suture of claim 10 packaged in a sterile and dry environment within a hermetically sealed and substantially moisture impervious container.

12. The method of closing a wound in living tissue which comprises approximating the wound tissue with an absorbable suture comprising of sterile, oriented fiber comprising an isomorphic copolyoxalate polymer consisting essentially of units of cyclic and linear oxalates and having the general formula



wherein each R is

$$-(CH_2)_n-A-(CH_2)_n- \quad \text{I}$$

or

$$-(CH_2)_{4+2n}- \quad \text{II}$$

with from about 5 to 95 mol percent of the R units being I; A is trans 1,4-cyclohexylene or p-phenylene, n is 1 or 2 and is the same for I and II, and x is the degree of polymerization resulting in a fiber forming polymer having a molecular weight greater than about 10,000.

13. The method of claim 12 wherein said fiber is a monofilament.

14. The method of claim 12 wherein said fiber is a multifilament.

15. The method of claim 14 wherein said multifilament fiber is a braid.

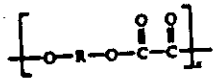
16. The method of claim 12 wherein n is 1 and A is trans 1,4-cyclohexylene.

17. The method of claim 12 wherein n is 2 and A is trans 1,4-cyclohexylene.

18. The method of claim 16 wherein units of formula I comprise from 40 to 75 mol percent of the R groups.

19. The method of claim 12 wherein A is p-phenylene.

20. A surgical prosthesis of a fabric manufactured at least in part from synthetic absorbable fibers comprising an isomorphic polyoxalate polymer consisting essentially of units of cyclic and linear oxalates and having the general formula:



wherein each R is

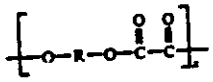
$$-(CH_2)_n-A-(CH_2)_n- \quad \text{I}$$

or

$$-(CH_2)_{4+2n}- \quad \text{II}$$

with from about 5 to 95 mol percent of the R units being I; A is trans 1,4-cyclohexylene or p-phenylene, n is 1 or 2 and is the same for I and II, and x is the degree of polymerization resulting in a fiber forming polymer having a molecular weight greater than about 10,000.

21. A surgical prosthesis of a solid surgical aid cast or machined from an absorbable polymer comprising an isomorphic polyoxalate polymer consisting essentially of units of cyclic and linear oxalates and having the general formula



wherein each R is

$$-(CH_2)_n-A-(CH_2)_n- \quad \text{I}$$

or

$$-(CH_2)_{4+2n}- \quad \text{II}$$

with from about 5 to 95 mol percent of the R units being I; A is trans 1,4-cyclohexylene or p-phenylene, n is 1 or 2, and is the same for I and II, and x is the degree of polymerization resulting in a fiber forming polymer having a molecular weight greater than about 10,000.

* * * * *

United States Patent [19]

Brennan et al.

[11] **Patent Number: 4,959,069**

[45] **Date of Patent: Sep. 25, 1990**

[54] **BRAIDED SURGICAL SUTURES**

[75] Inventors: **Karl W. Brennan**, Somerset; **Alison M. Skinner**, Long Valley, both of N.J.; **Gregory Weaver**, New Hope, Pa.

[73] Assignee: **Ethicon, Inc.**, Somerville, N.J.

[21] Appl. No.: **424,622**

[22] Filed: **Oct. 20, 1989**

[51] **Int. Cl.**[5] **A61B 17/00**

[52] **U.S. Cl.** **606/228**; 428/224; 87/7

[58] **Field of Search** 606/228, 229, 230, 231, 606/151; 138/123, 129; 139/317; 428/224

[56] **References Cited**

U.S. PATENT DOCUMENTS

3,035,583	5/1962	Hirsch et al.	606/223
3,054,406	9/1962	Usher	606/151
3,105,493	10/1963	Usher	606/231
3,187,752	6/1965	Glick	606/231
3,540,452	11/1970	Usher et al.	606/231
3,927,926	12/1975	Powell	24/115 N
3,949,755	4/1976	Vauquois	606/229
4,043,344	8/1977	Landi et al.	606/230
4,047,533	9/1977	Perciaccante et al.	606/230
4,275,117	6/1981	Crandall	428/373
4,510,934	4/1985	Batra	606/231
4,550,639	11/1985	Champlin	87/7
4,668,318	5/1987	Piccoli et al.	156/149
4,804,382	2/1989	Turnia et al.	623/1

Primary Examiner—Randall L. Green
Assistant Examiner—Gary Jackson

[57] **ABSTRACT**

A braided surgical suture is provided which in a first embodiment is woven in a spiral braid. The suture is braided by moving thread carriers from position to position around a circular path. As each carrier moves it moves from its present position to a succeeding position which is at least two positions removed from its present position. Such spiral braided sutures are advantageously produced without core filaments, providing benefits in strength, smoothness, pliability and cylindrical uniformity without the discontinuity of properties characteristic of conventionally braided cored sutures. In a second embodiment the suture is woven in a lattice braid, providing a plurality of distributed core passageways for individual core fibers.

17 Claims, 4 Drawing Sheets

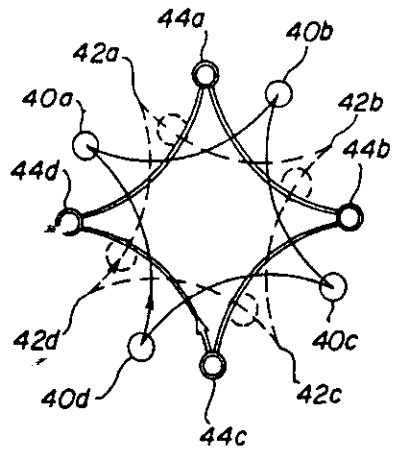


FIG-1a

PRIOR ART

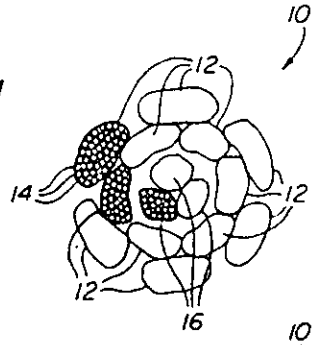


FIG-1b

PRIOR ART

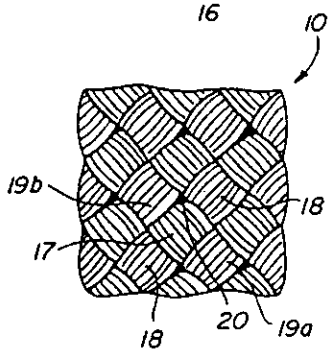


FIG-3

PRIOR ART

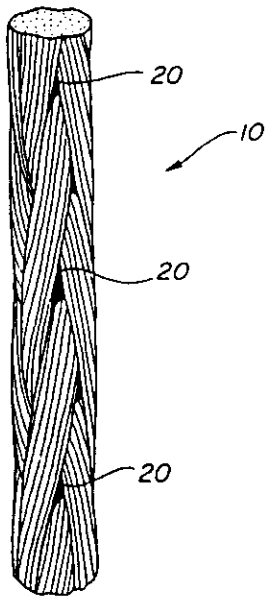


FIG-2 PRIOR ART

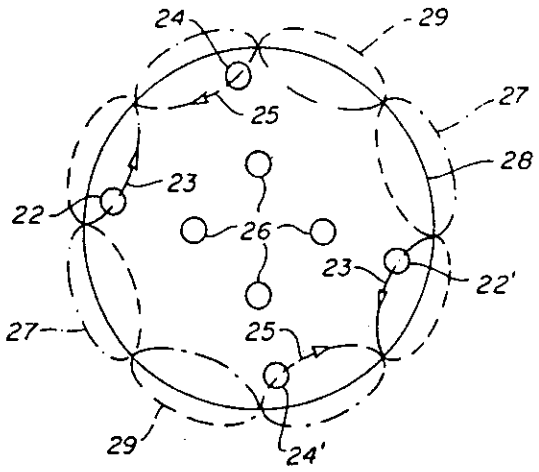


FIG-4a

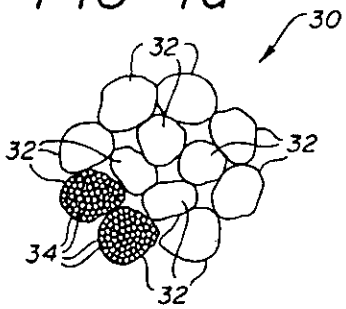


FIG-4b

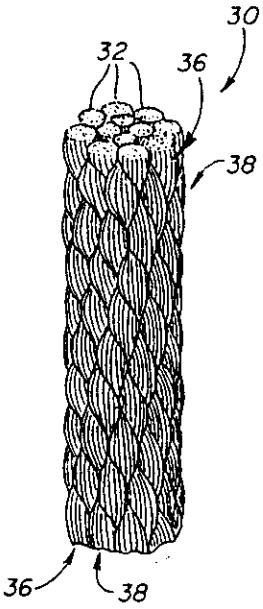


FIG-5a

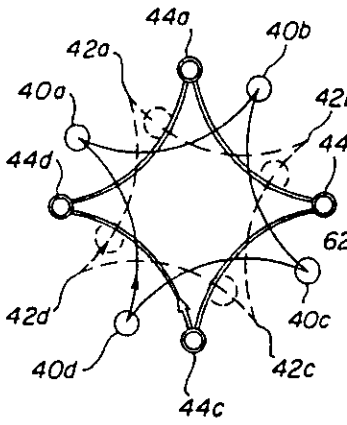


FIG-5b

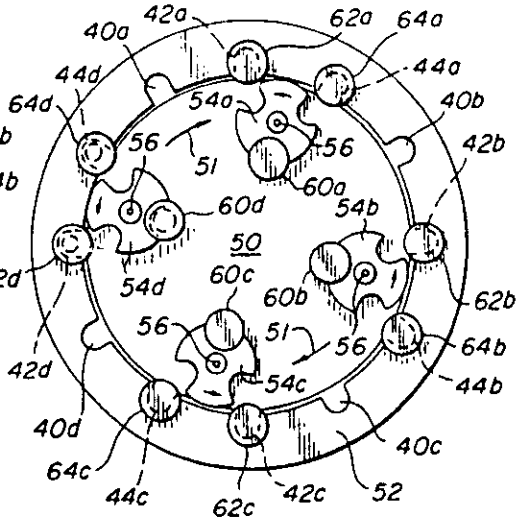


FIG-6a

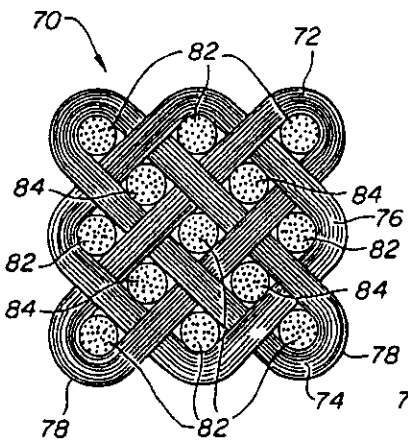


FIG-6b

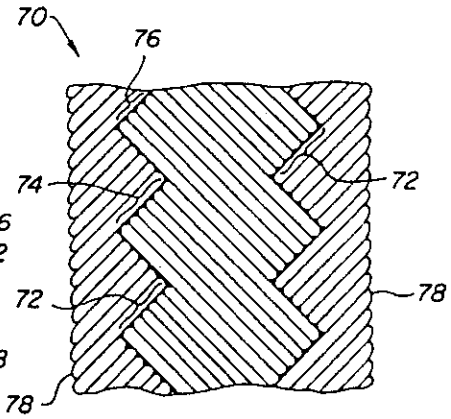
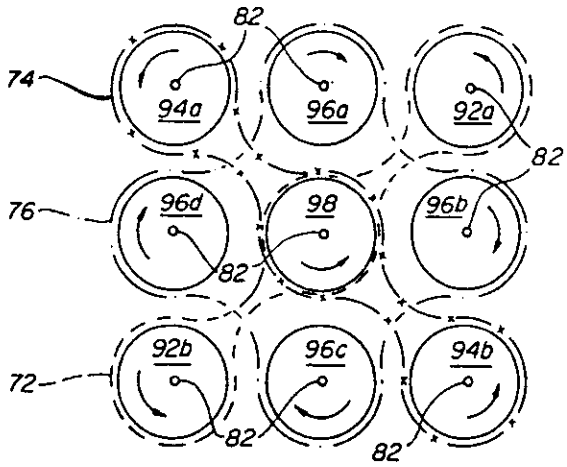


FIG-7



BRAIDED SURGICAL SUTURES

This invention relates to surgical sutures and, in particular, to braided surgical sutures which obviate the need for a central fiber core.

Surgical sutures may be manufactured in two general forms: monofilaments and multifilaments. Monofilament sutures are generally made of natural materials such as gut, or of extruded polymeric materials such as Nylon, polypropylene, or poly (p-dioxanone), and are highly regarded for their uniform, smooth construction and uniformly distributed tensile strength. However, monofilament sutures generally have the drawback of being fairly rigid and lacking pliability. Multifilament sutures consisting of a plurality of braided filaments of a fine gauge have been found to provide the characteristic of pliability which is often desired by surgeons Such braided sutures may be made of poly(lactide-co-glycolide), polyglycolide, polyester, or silk, for example. But since braided sutures often lack substantial tensile strength, the braided filaments are conventionally braided in a tubular sheath around a core of longitudinally extending threads. Such braided sheath sutures with central cores are shown in U.S. Pat. No. 3,187,752; 4,043,344; and 4,047,533, for example.

Braided sutures with central core threads have been found to exhibit certain disadvantages, however. One is that the tensile strength of the suture is not evenly distributed between the braided sheath and the central core threads. As a consequence, when these sutures are stretched, the sheath and the core will respond differently to the application of the tensile forces. The sheath will respond to the forces independently of the central core threads, causing the central threads to move longitudinally relative to the surrounding sheath. The core threads can also flatten and redistribute themselves within the sheath instead of maintaining the desired rounded cross-sectional shape of the suture. It would be desirable for such tensile forces to be more uniformly distributed throughout the suture, so that all of the fibers of the suture will respond in unison to the tensile forces without distortion of the normal shape of the suture.

Conventionally braided sutures can also feel rough to the touch, due to the changing crossing pattern of the braided filaments, and the interstices formed where the braided fibers overlap and cross each other. To minimize this tactile characteristic it is often necessary to further process the braided suture by heating and stretching the suture. Furthermore, such interstices can trap and retain moisture in a wicking fashion. Retained moisture has been found to be a source of undesired deterioration of sutures made of certain materials, such as absorbable sutures made of poly(lactide-co-glycolide) or polyglycolide, and can also lead to retention of sources of infection within the braid. It would be desirable to form braided sutures which are smoother to the touch, and which do not exhibit interstices or passageways which can trap and retain moisture prior to use of the sutures.

It would further be desirable for braided sutures to match or exceed the breaking strength characteristics of presently available braided sutures with central core threads.

In accordance with the principles of the present invention, a braided suture is provided in which the filaments or threads are braided in a spiral pattern. Sutures

braided in a spiral pattern have been found to be capable of maintaining a uniformly rounded cross-sectional shape, and to distribute tensile forces uniformly throughout the braided fibers. Spiral braided sutures also do not form the tube-like structure of the conventional braiding pattern, which eliminates the need for a central fiber core. Since spiral braiding results in an outer sheath pattern in which the braided threads are all flowing in the same direction, the suture is much smoother to the touch than the conventionally braided suture. The smoothly flowing braided configuration also does not provide interstices which can trap undesired moisture in the suture. Furthermore, the spiral braided suture has been found to be stronger, smoother to the touch, and much more pliable than the conventionally braided suture.

In accordance with a further aspect of the present invention a suture is provided which is formed by lattice braiding. The lattice braided suture exhibits a plurality of interwoven threads in a generally rectangular cross-sectional configuration. The lattice braid may be woven around a plurality of core threads distributed in the internal interstices of the lattice network and interlocked into position, unlike the central bundle of core threads of the conventionally braided suture. The lattice braided suture has been found to be superior to the conventionally braided core suture in that it does not exhibit "core pop", the tendency of the core filaments to break through the braided sheath as the suture is bent.

In the drawings:

FIGS. 1*a* and 1*b* illustrate diagrammatic cross-sectional and side views of a conventionally braided suture;

FIG. 2 illustrates the braiding pattern of a conventionally braided suture;

FIG. 3 is a drawing of an enlarged view of the outer sheath of a conventionally braided suture;

FIGS. 4*a* and 4*b* illustrate diagrammatic cross-sectional and side views of a spiral braided suture of the present invention;

FIG. 5*a* illustrates the braiding pattern of a spiral braided suture of the present invention;

FIG. 5*b* is a diagrammatic plan view of a mechanism used to braid a spiral braided suture of the present invention;

FIGS. 6*a* and 6*b* illustrate the braiding pattern and outside sheath of a lattice braided suture of the present invention; and

FIG. 7 is a diagrammatic plan view of a mechanism used to braid a lattice braided suture of the present invention.

Referring first to FIGURE 1*a*, a conventionally braided suture 10 is shown in diagrammatic cross-section. The suture 10 comprises a plurality of threads or carriers 12 which are interwoven to form the braided sheath. Each thread generally comprises a number of individual fibers 14. The braided threads 12 form a tubular sheath around the central core threads 16, which extend longitudinally through the tubular sheath. The sheath is braided using at least three threads, or a greater even number of threads, such as 4, 6, 8, etc. The core may comprise one or any greater number of threads. The suture 10 is shown in FIG. 1*a* to exhibit its desired cylindrical uniformity. However, it has been found that during handling, heating and stretching of the suture during manufacture the tubular sheath can distort to an oval or oblong shape, with the core threads

16 redistributed in the sheath in an irregular or linear configuration.

As a consequence of the structural independence of the braided sheath and the core threads, the sheath and core will unevenly distribute tensile forces among these two substructures when the suture is stretched, causing the two to move relative to each other. The relative movement of the two can result in the formation of spaces or pockets inside the sheath, between threads 16 of the core and the surrounding sheath. These spaces can entrap moisture through the mechanism of wicking, resulting in premature deterioration and weakening of the suture in in vivo use of the suture.

The conventionally braided suture is woven as indicated by the braiding pattern of FIG. 2, shown in a plan view. The individual threads of the braided sheath feed from spools mounted on carriers 22, 22' and 24, 24'. The carriers move around the closed circular loop 28, moving alternately inside and outside the loop 28 to form the braiding pattern. One or more carriers are continually following a serpentine path in a first direction around the loop, while the remaining carriers are following a serpentine path in the other direction. In the illustrated embodiment carriers 22, 22' are travelling around serpentine path 27 in a clockwise direction as indicated by directional arrows 23, and carriers 24, 24' are travelling around serpentine path 29 in a counterclockwise direction as indicated by arrows 25. Disposed within the center of the loop 28 are carriers 26 which dispense the core threads of the suture. Thus, the moving carriers 22, 22', 24, and 24' dispense threads which intertwine to form the braided sheath, and the sheath is formed around the centrally located core threads dispensed from carriers 26. The threads from all of the carriers in a constructed embodiment of FIG. 2 are dispensed upward with respect to the plane of the drawing, and the braided suture is taken up on a reel located above the plane of the drawing.

FIG. 1*b* is an illustration of the outside of the braided sheath of the suture 10 of FIG. 1*a*, showing the crossing pattern of the braided threads 12. Each thread is composed a number of individual fibers as indicated by the lines on each thread. Where each thread appears on the outside of the sheath it is seen to be orthogonally directed with respect to the thread it crosses over, the thread from beneath which it appears, and the thread it next crosses under. For instance, thread 17 is orthogonally directed with respect to thread 18 on either side of thread 17 where thread 17 crosses over thread 18. The thread 17 is also orthogonally directed with respect to thread 19*a* from beneath which it appears, and with respect to thread 19*b* which it next crosses under.

This orthogonal crossing relationship of the braided threads results in the formation of small interstices or voids 20 where the threads cross one another, as shown in FIG. 3, which shows a drawing reproduction of an enlarged photograph of a conventionally braided suture. These voids 20 can entrap moisture which can lead to premature deterioration of the suture, and can also entrap bacteria and other sources of infection causing complication of wound healing.

Referring now to FIG. 4*a*, a spiral braided suture 30 of the present invention is shown in diagrammatic cross-section. The braided suture 30 comprises a plurality of interwoven and interlocked threads 32, each of which may comprise a number of individual fibers 34. Due to the interlocking of the threads 32, no central passageway is formed in which moisture can become entrapped. The interlocking of the threads also causes the threads to move in unison as a continuous structure, thereby uniformly distributing tensile forces when the suture 30 is pulled or stretched.

The spiral pattern of the suture 30 is clearly shown in the outside view of the suture of FIG. 4*b*. The threads on the outside are seen to be aligned in a spiral pattern which ascends from the lower left to the upper right in the drawing as the outer threads precess around the outer surface of the suture. One spiraling set of threads is indicated between arrows 36, and another set is indicated between arrows 38. As the pattern spirals, individual threads on the outer surface are in a parallel orientation with respect to each other and with respect to the longitudinal length of the suture as they continually reappear in the spiral pattern.

With all threads aligned in the parallel, offset spiral pattern of FIG. 4*b*, it may be seen that there are no voids or interstices formed on the outside surface of the suture. This is due to the parallel orientation of the threads, as opposed to the orthogonally directed crossing pattern of the threads of the conventionally braided suture of FIGS. 1*b* and 3. The parallel orientation of the outer appearing threads also provides a smoother feel to the suture, since the hand will sense the continuous, longitudinal orientation of the parallel threads as it is run along the suture.

A spiral braided suture of the present invention is formed of four or more interwoven threads. Preferably at least nine threads are braided in groups of three, and a braiding pattern for a spiral braided suture of twelve threads, arranged in groups of four, is shown in FIG. 5*a*. In the illustrated pattern the carriers move sequentially in the same direction around the circular loop of carriers. As they move, each carrier moves from its present position to a succeeding position which is at least two positions removed from its present position. In the illustration of FIG. 5*a*, each carrier moves to the third succeeding position around the loop. The twelve carriers are grouped into three groups of four carriers each. In the first group, carriers move in unison between positions 42*a*, 42*b*, 42*c*, and 42*d*. The carrier at position 42*a* moves to position 42*b*, passing by positions 44*a* and 40*b* as it does so. As it moves, the carrier at position 42*b* is moving to position 42*c*, bypassing positions 44*b* and 40*c*. At the same time the carrier at position 42*c* is moving to position 42*d*, and the carrier at position 42*d* is moving to position 40*a*.

After these four carriers have moved to their new positions in unison, the carriers at positions 44*a*, 44*b*, 44*c*, and 44*b* move to their succeeding positions. Then the carriers at positions 40*b*, 40*c*, 40*d*, and 40*a* move to their succeeding positions. The sequence then repeats in the same fashion.

Apparatus for executing the spiral braiding pattern of FIG. 5*a* is diagrammatically shown in FIG. 5*b*. The apparatus comprises a rotating central platform 50 which is surrounded by an annular plate 52. The platform 50 rotates as indicated by arrows 51. Pivotally mounted on the platform 50 are four rotating carrier pickups 54*a*, 54*b*, 54*c*, and 54*d* which rotate about pivot points 56. Each pickup has a number of apertures which engage the carriers to move them to their succeeding positions, the number being chosen in correspondence with the number of positions to be bypassed as the carriers move in their braiding pattern. In the illustrated embodiment the number of apertures is three, enabling the carriers to bypass two positions each time they are

moved. The twelve carrier positions are delineated by rounded openings in the annular plate 52, four of which are indicated at 40*a*, 40*b*, 40*c*, and 40*d*. The carriers which carry spools of thread are indicated at 60, 62, and 64.

In operation pickup 54*a* will engage the carrier 60*a* at position 40*a*. As the central platform 50 rotates the pickup 54*a* simultaneously rotates to transfer the carrier 60*a* from position 40*a* to position 40*b*, which has just been vacated by carrier 60*b*. The carrier 60*a* is seen to bypass carriers 62*a* and 64*a* as it travels to its succeeding position 40*b*. As carrier 60*a* is transferred by pickup 54*a*, pickups 54*b*, 54*c*, and 54*d* simultaneously are transferring carriers 60*b*, 60*c*, and 60*d* to their succeeding positions.

As the pickup 54*a* is about to deposit the carrier 60*a* at position 40*b*, the pickup engages carrier 64*a* to begin transferring that carrier to its succeeding position. The carriers 64*b*, 64*c*, and 64*d* are similarly engaged simultaneously by the other three pickuPs. After the carriers 60*a*, 60*b*, 60*c*, and 60*d* have been deposited at their new positions and the carriers 64*a*, 64*b*, 64*c*, and 64*d* are enroute to their succeeding positions, the pickups engage the carriers 62*a*, 62*b*, 62*c*, and 62*d* for transfer. As this sequence of carrier transfer continues, threads from the spools on the carriers are dispensed upward with respect to the plane of the drawing and the braided suture is taken up on a reel located above the apparatus.

A lattice braided suture 70 of the present invention is shown in FIG. 6*a*, which schematically illustrates the structure of the lattice braid. In FIG. 6*a*, three or more threads are braided in a lattice pattern. One thread or group of threads traverses the path 72, a loop extending from the upper right to the lower left of the drawing. As the carrier or carriers dispensing thread on path 72 move around this path, they alternately cross over and under the paths of the other threads that they encounter, the crossing pattern being determined by the times and locations of travel of the respective carriers. In a similar fashion a second carrier or carriers dispensing thread traverse a path 74 from the lower right to the upper left of the pattern. Like the first path, the thread dispensed from carriers travelling this path alternately crosses over and under the other paths it encounters. A third path 76 travels around the intersection of the first path 72 and the second path 74. Like the first two paths, the thread dispensed from the carrier or carriers traversing path 76 alternately crosses over and under the threads of the other paths it encounters.

The lattice braid of FIG. 6*a* is seen to exhibit a generally square shape in cross-section with rounded corners. While the lattice braided suture has been found to provide less tensile strength than the spiral braided suture, the lattice braided suture can be strengthened by the inclusion of individual core threads running longitudinally through the interlocking lattice. A number of core threads may be located at the positions indicated at 82 in the lattice, at the positions indicated at 84, or both. This uniform distribution of core threads throughout the lattice, which results in secure capture of the individual threads within the loops of the lattice, has been found to provide a uniform distribution of tensile forces throughout the suture.

The outside of the lattice braided suture 70 is illustratively shown in FIG. 6*b*. The outer threads of the lattice are seen to be distributed in an angularly offset, generally parallel configuration. The drawing shows the generally parallel alignment of threads 72, 74, and 76 on the outside of the suture, forming a substantially smooth, longitudinally extending outer thread surface on each side of the square configuration. The rounded corners 78, shown on each side of the drawing, are also seen to smoothly extend along the length of the suture.

Apparatus for braiding the lattice braided suture of FIG. 6*a* is schematically shown in FIG. 7. The apparatus includes a plurality of rotating discs which transfer the carriers around and along their intended paths of travel. In a preferred embodiment there are three carriers traversing each path. Extending through the center of each rotating disc is a core thread 82, each of which becomes engaged in the lattice loops formed around its respective disc. The path 72 is traversed by carriers which rotate around disc 92*a* and are then transferred to the central disc 98. Each carrier travels halfway around disc 98 and is then transferred to disc 92*b*. The carriers travel around disc 92*b* and back to the central disc 98. After travelling around the other side of disc 98 each carrier is transferred back to rotating disc 92*a* and its starting point.

In a similar manner a second group of carriers on the path 74 travel around disc 94*a* and are transferred to the central disc 98. After travelling halfway around disc 98 each carrier is transferred to disc 94*b*. Each carrier travels around the rotating disc 94*b*, back to the other side of the central disc 98, and is returned to disc 94*a* and its starting point.

The third path 76 passes around rotating discs 96*a*, 96*b*, 96*c*, and 96*d*. The carriers which travel this path 76 pass around three-quarters of each disc before being transferred to the succeeding disc in the loop. As each carrier traverses the path 76 it is seen to pass inside the end discs of the other two paths 72 and 74, thereby enclosing the intersection of these two paths at the central disc 98.

The apparatus of FIG. 7 may be operated with a plurality of carriers travelling each path simultaneously. For instance, the apparatus may be operated with three carriers on each path to form a lattice braid of 9 threads. Alternatively each path may include 4 carriers for a total of 12 braided threads. As a third example, the apparatus may operate with 6 carriers on each path for a total of 18 threads in the braided suture.

Spiral braided sutures of the present invention can be expected to provide a 20% improvement in smoothness over conventionally braided sutures, a 20% improvement in pliability, and a 50% improvement in cylindrical uniformity. The improvement in smoothness is due to the parallel alignment of the suture threads on the outside of the spiral braided suture. The improvement is pliability is due to the thread crossovers of the spiral braid, which enhances fiber mobility; the individual threads in the spiral braided suture will easily move relative to each other as the suture is bent. And since there is no core to become misaligned or misshapen, cylindrical uniformity is improved.

Improvements in breaking strength can also be expected for the spiral braided suture. In a test of breaking strength remaining (BSR) after 21 days of in vivo use of an absorbable suture of conventional braid, typically 40-50% of the breaking strength remains. A 15-20% improvement in BSR can be expected in use of a spiral braided suture of the present invention under the same conditions.

The lattice braided suture provides the capability of producing a high quality composite suture, in which advantage is taken of the different characteristics of one

type of material for the braid and another type of material for the core threads. As discussed above, the lattice braided suture is substantially more immune to the problem of core pop than the conventionally braided suture, since the core threads are distributed throughout the structure of the braid and are not positioned in a single central location. Both the spiral and lattice braided sutures have been found to exhibit less surface area exposed to ambient conditions, and hence less exposure to moisture, than conventionally braided sutures.

What is claimed is:

1. A braided surgical suture in which a plurality of surgically compatible filaments are woven in a spiral braid by moving filament dispensers to different positions around a closed loop, wherein an individual dispenser in the loop is moved from its current position to a succeeding position which is at least two positions removed from said current position.

2. The braided surgical suture of claim 1, wherein the number of filaments is at least nine.

3. The braided surgical suture of claim 2, wherein said filament dispensers move around said loop in the same direction.

4. The braided surgical suture of claim 3, wherein said filament dispensers are organized in three uniformly distributed groups around said loop and the dispensers in each group move around said loop in unison.

5. The braided surgical suture of claim 3, wherein the number of filaments is twelve and wherein an individual dispenser in the loop is moved from its current position to a succeeding position which is three positions removed from said current position.

6. The braided surgical suture of claim 1, wherein the portions of said filaments which are visible on the outside of said braided suture are oriented substantially parallel to each other and are distributed in patterns which spiral around the outside of said suture.

7. The braided surgical suture of claim 1, wherein said surgically compatible filaments are woven in a spiral braid without any central, longitudinally extending core filaments.

8. A braided surgical suture in which a plurality of surgically compatible filaments are woven in a lattice braid by moving filament dispensers in three closed loop paths, a first and second of said paths being generally oblong and crossing over each other at a central intersection, and the third of said paths passing through the ends of said first and second paths outside said central intersection.

9. The braided surgical suture of claim 8, wherein said suture in cross-section exhibits a generally rectangular shape, with filaments traversing said ends of said first and second paths being located at the corners of said rectangular shape.

10. The braided surgical suture of claim 8, wherein each moving filament dispenser alternately passes over then under the filaments dispensed on the paths it intersects.

11. The braided surgical suture of claim 10, wherein there are at least three filament dispensers traversing each of said paths.

12. The braided surgical suture of claim 8, wherein there are formed a plurality of core fillament passageways located adjacent to the points of intersection of two or more of said paths.

13. The braided surgical suture of claim 12, further comprising at least four core filaments located in ones of said passageways.

14. The braided surgical suture of claim 13, wherein said core filaments are symmetrically distributed with respect to said point of intersection.

15. The braided surgical suture of claim 13, wherein said core filaments are made of a different surgically compatible material than that of said woven filaments.

16. The braided surgical suture of claim 12, further comprising at least nine core filaments located in ones of said passageways.

17. The braided surgical suture of claim 12, further comprising at least thirteen core filaments located in ones of said passageways.

* * * * *

United States Patent [19]

Silvestrini

[11] Patent Number: 4,979,956

[45] Date of Patent: Dec. 25, 1990

[54] DEVICE AND METHOD FOR TENDON AND LIGAMENT REPAIR

[75] Inventor: Thomas A. Silvestrini, East Lyme, Conn.

[73] Assignee: Pfizer Hospital Products Group, Inc., New York, N.Y.

[21] Appl. No.: 378,437

[22] Filed: Jul. 10, 1989

Related U.S. Application Data

[63] Continuation of Ser. No. 115,087, Oct. 30, 1987, abandoned.

[51] Int. Cl.[5] A61F 2/06

[52] U.S. Cl. 623/13

[58] Field of Search 623/11, 12, 16, 13, 623/18

[56] References Cited

U.S. PATENT DOCUMENTS

3,124,136	3/1964	Usher	623/11
3,176,316	4/1965	Bodell	3/1
3,545,008	12/1970	Bader, Jr.	623/13
3,646,615	3/1972	Ness	623/13
3,805,300	4/1974	Tascon-Alonso	623/13
3,987,497	10/1976	Stoy et al.	623/13
3,992,725	11/1976	Homsy	623/16
4,469,101	9/1984	Coleman et al.	128/334 R
4,501,029	2/1985	McMinn	3/1
4,512,038	4/1985	Alexander et al.	623/13
4,585,458	4/1986	Kurland	623/13
4,610,688	9/1986	Silvestrini et al.	623/1

FOREIGN PATENT DOCUMENTS

197703	1/1976	U.S.S.R.	623/16
197708	8/1977	U.S.S.R.	128/334 R

OTHER PUBLICATIONS

K. Cieslik et al., Early Mechanical Strength of Digital Flexor Tendon Sutures, Handchirurgie, vol. 18, Nov. 1986, pp. 347-350.

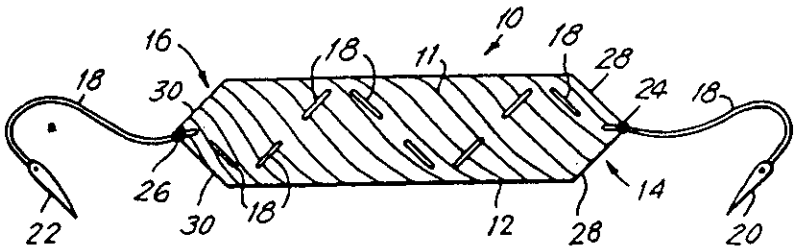
Primary Examiner—David J. Isabella

Attorney, Agent, or Firm—Peter C. Richardson; Lawrence C. Akers; John L. LaPierre

[57] ABSTRACT

A device, suitable for use in repairing a lacerated or severed tendon, particularly a hand flexor tendon, having a flat band body with opposite ends of the body designed to anchor connecting sutures. The device also finds applicability in the repair of lacerated or severed ligaments. Also disclosed is a method of repairing a severed tendon by implanting a flat band device suturing together the device and the tendon to effect an anastomosis along approximated ends of the severed tendon. Further disclosed is a method of repairing a lacerated or severed ligament.

28 Claims, 2 Drawing Sheets



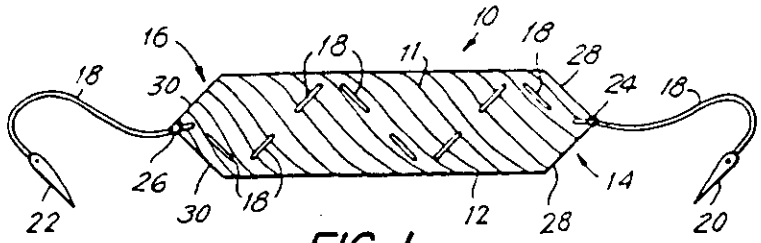


FIG. 1

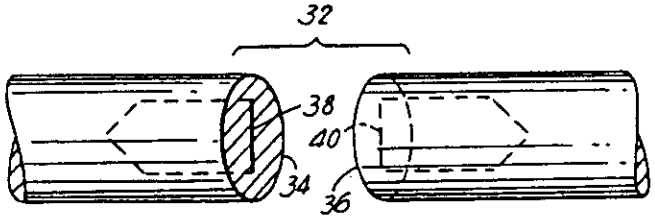


FIG. 2

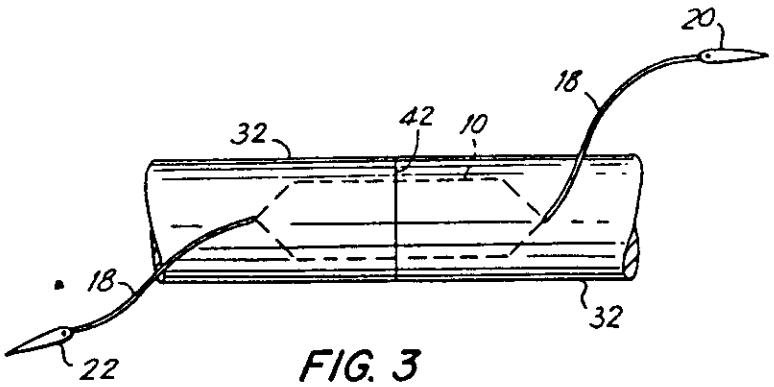


FIG. 3

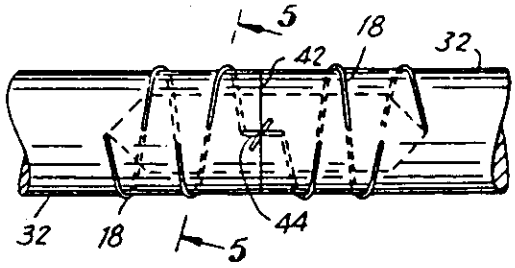


FIG. 4

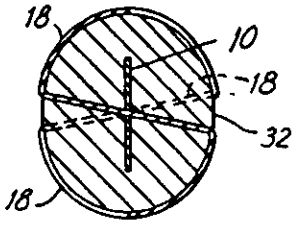


FIG. 5

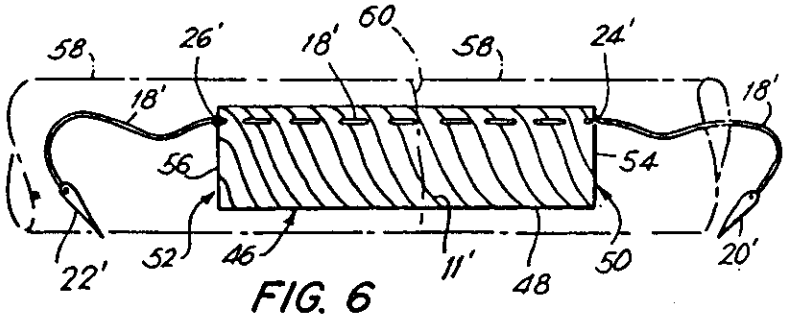


FIG. 6

DEVICE AND METHOD FOR TENDON AND LIGAMENT REPAIR

This is a continuation, of application Ser. No. 115,087, filed on Oct. 30, 1987, now abandoned.

BACKGROUND OF THE INVENTION

The present invention relates to a device for repairing severed or lacerated tendons and ligaments and, more particularly, the invention relates to a device having a flat band body constructed of resilient synthetic textile fiber and capable of receiving a suture element secured thereto at opposite ends of the body. Also contemplated by the invention are methods of anastomosing ends of severed or lacerated tendons and ligaments along an interface of approximated ends, by placing a device intratendonously, in the case of a tendon, or in juxtaposition, in the case of a ligament, bridging or spanning approximated ends, and suturing together either the tendon and the device or the ligament and the device. The objective is to provide a device and method for restoring tendons and ligaments, as nearly as possible, to their pre-damaged condition.

The successful repair of tendons, particularly hand flexor tendons, has been a problem for surgeons for many years. The past and current approach most commonly used by surgeons to achieve tendon repair is to anastomose severed tendons by using one of a variety of suturing techniques. A number of such techniques are commonly known and referred to as Bunnell, Kessler, Klienert, Tsuge and Becker, to name but a few. These techniques, while useful, are not entirely satisfactory because they allow surgeons to achieve successful repairs in only about 70% of the patients treated. Therefore, in view of the history of suture techniques which have been proposed and implemented from time to time by surgeons without any real improvement in repair strength or surgical result, the need for an improved device and method of anastomosis were clearly evident.

In addition to the foregoing suturing techniques most often used in tendon repair, in an effort to overcome the deficiencies encountered in the straight suturing approach, other devices and approaches have recently been tried to effect tendon repair. A typical device encountered might be one like that disclosed in U.S. Pat. No. 4,469,101. The teaching embodied in this patent specifies a structure having an open network or mesh of helically formed members to define a hollow tubular device wherein opposing ends of a lacerated tendon are introduced and brought into contact within the tube. The opposite ends of the tube are then sutured to the outer tendon wall and the contacting tendon ends are allowed to heal. Another device typically encountered in tendon repair might be one like that disclosed in U.S. Pat No. 4,501,029. This patent provides a continuous solid wall tubular device having in communication therewith a number of transversely extending passages. The tube is inserted between a replacement tendon and the tendon sheath. After blood supply from the sheath to the replacement tendon is established through the tubular passages, free movement of the tendon is established within the sheath. A third device encountered might be the plastic prosthetic tendon disclosed in U.S. Pat. No. 3,176,316. This patent provides a prosthesis having a solid central segment and hollow tubular ends comprising a mesh network wherein ends of a tendon are introduced and the prosthesis is sutured to the tendon.

There are certain disadvantages associated with each of the aforementioned tendon repair techniques and devices which the present inventive device and method either overcome or substantially lessen. Specifically, through the use of suturing techniques alone, irritations are minimized since sutures are buried inside the endotendon, but the strength of the anastomosis is not strong enough to allow aggressive mobility during healing. Consequently, there often occurs dehiscence of the suture leading to separation of approximated tendon ends, tissue ingrowth and slow or incomplete tendon healing. Inherent in the tubular mesh devices which are sutured to the tendon at ends of the devices is the exposure of a large amount of synthetic material on the outside of the epitenon which can cause excessive irritations. These irritations frequenty lead to adhesions between the injured tendon and the tendon surrounding which leads to retarded healing. The inventive device offers a minimum of irritation since it is substantially buried inside the endotendon, yet it offers higher strength of the anastomosed tendon compared to repairs using sutures. Lastly, the present device is one of structural simplicity which avoids both the complex geometry presented in the solid wall tubular device having a series of selectively positioned blood conveying passageways and the need to precisely locate such a prosthesis in the body to assure an adequate blood supply to the replacement tendon.

It should be understood that, while much of the foregoing discussion is directed toward tendon repair, the teachings encountered are also generally applicable to the repair of damaged ligaments. Clearly, there exists a need for a repair device which fosters superior mechanical repair properties and better healing characteristics than is currently found in the relevant surgical field. The present inventive device and method satisfies the need and, hence, advances the art field of tendon and ligament repair.

SUMMARY OF THE INVENTION

The present invention relates to a device used for repairing severed connective tissue of tendons and ligaments by approximating ends of the severed tissue and comprises an elongated body portion having a flat band structure with the body portion at opposite ends adapted to be connected to at least one needle bearing suture. The body structure may be a non-woven fabric, a composite reinforced with chopped fiber, a polymer sheet or a fabric which can be selected from a class of warp knits, weaves, nets and braids. The preferred braided fabric would be a triaxial braid or a flat band triaxial tube having either a monocomponent or bicomponent fiber element selected from a polymeric grouping and may include an elastomeric component. The preferred polymer for a monocomponent device body would be polyethylene terepthalate while for a bicomponent device the preferred polymers for the device body would be polyethylene terepthalate and polyester/polyether block copolymer. A suture or sutures may be lock stitched to opposite ends of the device body and may be incorporated into the body structure axially in either a longitudinal direction or in a bias direction. Additionally, a suture or sutures may be sewn into the body. The device body and associated suture or sutures may be covered with one or more gel coatings selected from a class of hydrogels with a preferred

coating being crosslinked calcium alginate. The body portion may assume a number of shapes but either a rectangle or a polygon, having ends tapered substantially to a point, is preferred. The ends of the body portion are preferably sealed to maintain edge integrity.

Also contemplated within the scope of the present invention are methods for repairing severed connective tissue of tendons and ligaments utilizing the inventive device heretofore described. Specifically, one method comprises the steps of creating a slot in the tissue of each opposing end of a severed tendon, where severance occured, inserting a first end of the device into one of the incised slots, inserting a second end of the device into the other of the incised slots, approximating opposing ends of the severed tissue, enclosing the device and therewithin bridging the ends, and suturing the tendon and the device together, passing sutures through the tendon and the implanted device along at least a portion of the length of the device, to anastomose the tendon along approximated ends of the severed connective tissue. A second method, relating to the repair of severed connective tissue of a ligament, comprises the steps of providing at least one inventive device, approximating opposing ends of the severed tissue, juxtaposing the ligament and the device with the device spanning approximated ends, and suturing the ligament and the device together, passing sutures through the ligament and the juxtaposed device along at least a portion of the length of the device, to anastomose the ligament along approximated ends of the severed connective tissue. In each of the methods, suturing will span at least the approximated ends and, preferably, suturing will be performed along substantially the entire length of the device.

The various features of novelty which characterize the invention are pointed out with particularity in the claims annexed to and forming a part of this disclosure. For a better understanding of the invention, its operating advantages and specific results obtained by its use, reference should be made to the corresponding drawings and descriptive matter in which there is illustrated and described typical embodiments of the invention.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is an enlarged schematic representation of a tendon and ligament repair device, in accordance with the principles of the present invention, illustrating a flat band triaxial braid fabric structure having a single bias suture incorporated into the fabric body with the suture lock stitched to the body at opposite ends of the body.

FIG. 2 schematically illustrates a severed tendon, drawn at reduced scale, with slots incised in the tendon ends, before implantation of the repair device.

FIG. 3 is similar to FIG. 2, but with tendon ends approximated, and schematically illustrates the tendon repair device of FIG. 1 located within the endotendon prior to suturing.

FIG. 4 is similar to FIG. 3 and illustrates a completed repair showing suture penetration of both tendon and fabric body uniting tendon and device.

FIG. 5 is a cross-sectional view taken along line 5—5 of FIG. 4.

FIG. 6 is an enlarged schematic alternate embodiment of the invention showing in phantom a ligament with approximated ends and a flat band triaxial braid fabric structure, in place but prior to suturing, with a single axial suture incorporated into the fabric body with the suture lock stitched to the body at opposite ends thereof.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

The description herein presented refers to the accompanying drawings in which like reference numerals refer to like parts throughout the several views, and in which, referring to FIG. 1, there is illustrated a repair device 10 of the present invention. The device has an elongated body portion 12 of flat band structural configuration, preferably a triaxial braid with the braid schematically designated as 11, and at a first end 14 and at a second end 16 a suture 18, having needles 20 and 22 at opposite ends, is connected or anchored to the body ends by a locking stitches 24 and 26. It should be understood that many types of knots or locking stitches, such as a double throw suture locking stitch, would be suitable to anchor the suture to the body portion. Locking the suture to the body could be accomplished at any time, as desired. The device shown in FIG. 1, it should be remembered, is a schematic representation and, therein depicted, the device has a flat band triaxial braid fabric structure with a single bias suture braided into the fabric body. It should also be understood that more than one suture could be attached to or incorporated into the fabric body and locked to the body ends. Furthermore, a suture or sutures could be sewn or stitched to the body along the body length instead of being braided into the body. In the preferred form of the device, the lock stitching of the suture to the ends of the braid body prevents the braid structure from bunching during insertion into tissue. The stitching also serves to center the suture pull of the device, thereby easing the insertion of the device into connective tissue. Also contemplated within the scope of the invention is a suture or sutures not incorporated into the fabric body per se but merely locked to one or both of the body ends. The ends 14 and 16 may be sealed along edges 28 and 30 to maintain edge integrity. Edge sealing may be accomplished using an ultrasonic sealing process or other means of heat treatment to keep edges from unraveling or separating.

The device body portion may be structurally configured as a non-woven fabric, a polymer reinforced with chopped fiber, a polymer sheet, a warp knit, a weave, a net or a braid. The construction of the desired flat band fabric into any one of these body portion structural configurations would be within the skill of those who manufacture textile products. A preferred structure would be a braid, preferably a triaxial braid. A flat band or flattened triaxial tube is within the scope of the invention. A triaxially-braided fabric, such as the ones schematically depicted in FIG. 1 and FIG. 6, and the methods of manufacturing them in different configurations, namely, flat bands, flat tubes, tubes, patches and strips, to name but a few, are well known to those skilled in the art of manufacturing braided polymeric articles. The triaxial braid may consist of a monocomponent fiber selected from a group of polymers consisting of polyethylene terepthalate, polyethylene, polypropylene, polyaramid, polyamide, polyetheretherketone, polyester/polyether block copolymer, liquid crystal polymeric fiber, nylon and carbon. The preferred polymer would be polyethylene terepthalate. The triaxial braid may also have a bicomponent fiber makeup with its components selected from the same polymer grouping. One of the components of the bicomponent

braid should be elastomeric with the preferred elastomer being polyester/polyether block copolymer. The preferred bicomponent braid comprises a first component of polyethylene terepthalate and a second component of polyester/polyether block copolymer.

The device may be coated to improve the ease of surgical installation and to minimize irritation to tissue during healing. The suture or sutures could also be coated to minimize adhesions formed during healing. The coating could be a gel, specifically a hydrogel, selected from the group consisting of sodium alginate, hyaluronic acid, crosslinked hyaluronic acid, crosslinked calcium alginate and a calcium alginate crosslinked hyaluronic acid mixture. The preferred lubricious coating for the device and sutures is crosslinked calcium alginate.

The device body as shown in FIG. 1 defines a polygon having opposed longitudinal ends each tapering to a point with the points, preferably, lying along the central longitudinal axis. The body may, however, as is shown in FIG. 6, take a rectangular shape. Other flat band structural shapes would be suitable and are within the scope of the present invention.

Turning to FIG. 2 through FIG. 5, in FIG. 2 there is shown severed connective tissue of a tendon 32 having separated ends 34 and 36. In each end 34 and 36, slots 38 and 40 have been incised within the endotendon using a suitable blade or cutting device (not shown). Each slot 38 and 40 will preferably be configured to conform substantially to one half the size of the repair device 10. FIG. 3 shows device 10 located within slots 38 and 40, suture 18 at opposite ends 14 and 16 of device 10 passing through tendon 32, and separated tissue ends 34 and 36 approximated as shown at 42. Device 10 is closed within the approximated tissue, bridging ends 34 and 36 which are in contact along joint 42. FIG. 4 and FIG. 5 depict a completed repair wherein the tendon and the device have been sutured together and the suture ends tied at 44. Suturing of the device into the tendon can be accomplished in many different ways. Thus, the device does not restrict the personal suturing preference of different surgeons. Anastomosis of the tendon will occur along approximated ends at 42. Suturing should span at least the approximated ends and, preferably suturing should be performed along substantially the entire length of the implanted device 10.

Turning to FIG. 6, there is schematically shown an alternate embodiment of the invention. Here depicted is a rectangular flat band repair device 46 having a triaxially braided fabric structure 11' and a suture 18', bearing needles 20' and 22', incorporated into elongated body portion 48 and axially oriented in a longitudinal direction. At first and second ends 50 and 52, suture 18' is affixed to the body ends by locking stitches 24' and 26'. As aforementioned in respect to the device 10, many types of knots or locking stitches would be suitable to affix the suture to the body portion and stitching could be accomplished when desired, namely, at time of manufacture or by a surgeon prior to device use. Lock stitching would be particularly useful, in addition to ease in installation, that is, prevention of fabric bunching, to keep the suture from being pulled through the fabric. More than one suture could be used and attached to or incorporated into the body fabric. Additionally, a suture might be sewn to the body along the length of the body rather than being braided into the body. The ends 50 and 52 may be sealed along edges 54 and 56 to maintain edge integrity, as in the case of device 10. All of the other structural features associated with device 10 are equally suitable for device 46.

In FIG. 6, device 46 is shown to be particularly useful in the repair of severed connective tissue of a ligament, illustrated in phantom and designated as 58. It should be understood, however, that a device of rectangular configuration would be equally useful in tendon repair and slots 38 and 40, as shown in FIG. 2, could assume a rectangular shape. Likewise, device 10 would be equally suitable in the repair of a ligament. Device 46, as provided in FIG. 6, is shown positioned alongside ligament 58 having severed ends approximated at 60. The device spans the approximated ends. It should be understood that more than one device could be used for the repair. While a completed repair is not shown in FIG. 6, a suturing technique like that shown in FIG. 4, and other techniques described in respect thereto, could be used to suture together ligament 58 and device 46. Anastomosis of the ligament will occur along approximated tissue ends at 60. Suturing should span at least the approximated ends and, preferably, suturing should be performed along substantially the entire length of device 46. In each of the repair techniques, namely, tendon and ligament, devices 10 and 46 are biocompatible and can be made from permanent, non-body absorbable materials, or from resorbable materials.

As heretofore mentioned, braiding can be accomplished using known technology and the inventive device can be manufactured using existing braiding machines modified to incorporate longitudinal fibers into the braided structures. By way of example, and not to be construed as limiting the invention, a 0.07 inch wide monocomponent polyethylene terepthalate device 10 can be braided on a 32-carrier triaxial braider using 70 denier white polyethylene terepthalate type 52 multifilament yarns and a single green 4-0 polyethylene terepthalate suture. The finished product is composed of 31 polyethylene terepthalate yarns and one 4-0 polyethylene terepthalate suture on the bias and 16 polyethylene terepthalate yarns on the longitudinal axis. In another example, a 0.07 inch wide bicomponent device 10 can be braided on a 24-carrier triaxial braider using 220 denier polyester/polyether block copolymer monofilaments, 70 denier white polyethylene terepthalate type 52 multifilament yarns, and a single green 4-0 polyethylene terepthalate suture. The finished construction is composed of 23 polyethylene terepthalate yarns and one 4-0 polyethylene terepthalate suture on the bias, and 12 polyester/polyether block copolymer fibers on the longitudinal axis. It should be understood that wider or narrower devices could be manufactured. The device is made from safe materials that surgeons are comfortable implanting and the device can easily be made in a variety of sizes to address different soft tissue repair situations. Device needles could be swaged onto the suture ends of affixed by other suitable means. Laboratory testing of a repair device used to anastomose explanted canine and bovine tendon has demonstrated that the initial strength of the repair junction is approximately twice the strength of tendon repairs made using conventional suturing techniques.

While in accordance with provisions of the statutes there is described herein specific embodiments of the invention, those skilled in the art will understand that changes may be made in the form of the invention covered by the claims appended hereto without departing from the scope and spirit thereof, and that certain features of the invention may sometimes be used to an

advantage without corresponding use of the other features.

I claim:

1. A device for use in repairing severed connective tissue of tendons and ligaments by approximating severed ends of said tissue bringing said tissue ends into abutment comprising an elongated body having a flat band structure, said body being sized and configured for enclosure within said abutting tissue ends, and with said body at first and second opposed non-bifurcated ends adapted to be connected to at least one needle bearing suture, with said suture being incorporated into said body substantially the length thereof, said suture being oriented in a bias direction.

2. The device according to claim 1 wherein said structure is a non-woven fabric.

3. The device according to claim 1 wherein said structure is a polymer reinforced with chopped fiber.

4. The device according to claim 1 wherein said structure is a polymer sheet.

5. The device according to claim 1 wherein said structure is a fabric selected from the group consisting of warp knits, weaves, nets and braids.

6. The device according to claim 5 wherein said fabric is a braid.

7. The device according to claim 6 wherein said braid is a triaxial braid.

8. The device according to claim 7, wherein said braid comprises a monocomponent fiber forming element.

9. The device according to claim 8 wherein said element is a polymer selected from the group consisting of polyethylene terepthalate, polyethylene, polypropylene, polyaramid, polyamide, polyetherether ketone, polyester/polyether block copolymer, liquid crystal polymeric fibers, nylon and carbon.

10. The device according to claim 9 wherein said polymer is preferably polyethylene terepthalate.

11. The device according to claim 7 wherein said braid comprises a bicomponent fiber forming element.

12. The device according to claim 11 wherein said element is a plurality of polymers selected from the group consisting of polyethylene terepthalate, polyethylene, polypropylene, polyaramid, polyamide, polyetherether ketone, polyester/polyether block copolymer, liquid crystal polymeric fibers, nylon and carbon.

13. The device according to claim 12 wherein at least one of said polymers is elastomeric.

14. The device according to claim 13 wherein said elastomeric polymer is preferably polyester/polyether block copolymer.

15. The device according to claim 12 wherein said polymers are preferably polyethylene terepthalate and polyester/polyether block copolymer.

16. The device according to claim 1 wherein said suture is lock stitched to said ends.

17. The device according to claim 1 wherein said suture is sewn into said body.

18. The device according to claim 17 wherein said suture is axially oriented in a longitudinal direction.

19. The device according to claim 1 wherein said body is covered with a gel coating.

20. The device according to claim 19 wherein said body and said suture are covered with a gel coating.

21. The device according to claim 20 wherein said coating is a hydrogel selected from the group consisting of sodium alginate, hyaluronic acid, crosslinked hyaluronic acid, crosslinked calcium alginate and a calcium alginate crosslinked hyaluronic acid mixture.

22. The device according to claim 21 wherein said hydrogel is preferably crosslinked calcium alginate.

23. The device according to claim 1 wherein said body defines a polygon.

24. The device according to claim 23 wherein said polygon is a rectangle.

25. The device according to claim 23 wherein at least one of said ends of said body terminates substantially in a point.

26. The device according to claim 25 wherein said point lies along a central axis of said body.

27. The device according to claim 1 wherein said ends are sealed proximate end edges to maintain edge integrity.

28. The device according to claim 1 wherein said body is a flat band triaxial tube.

* * * * *

As Originally Filed

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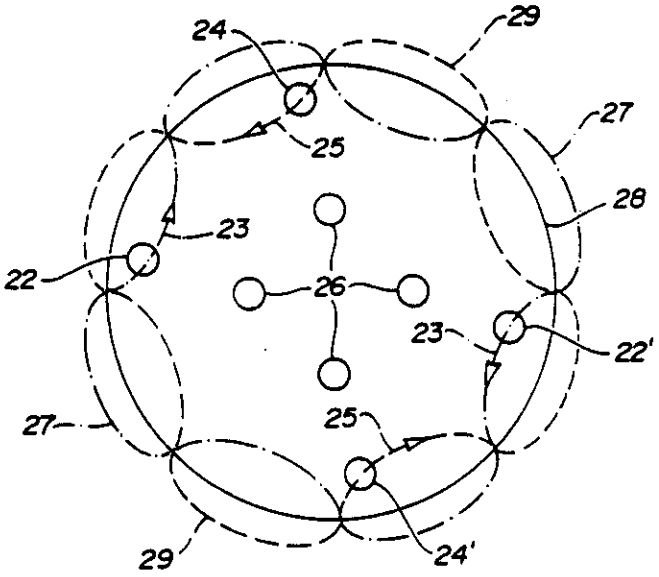
FIG-1



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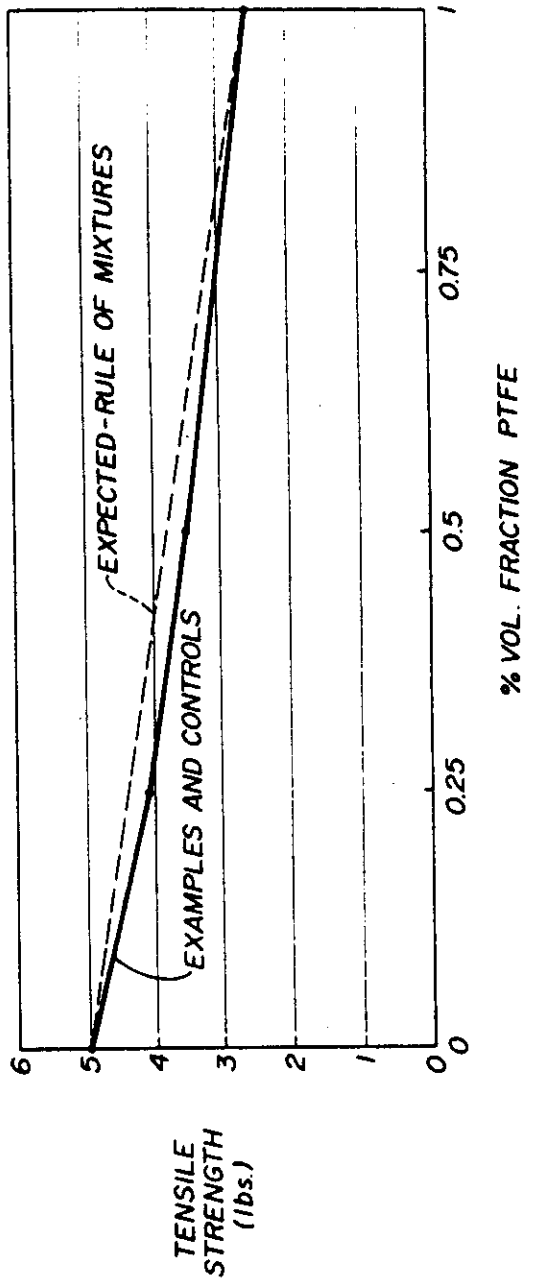

FIG-2

As Originally Filed

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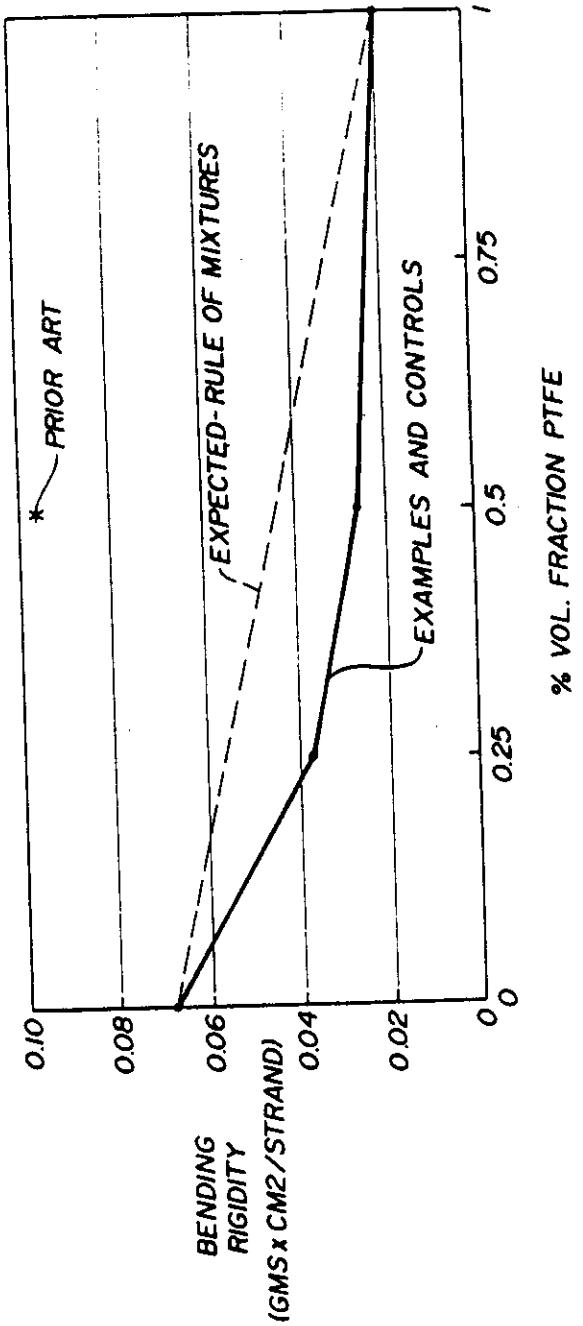
FIG-3



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   d 11 16

11.  4,959,069, Sep. 25, 1990, Braided surgical sutures; Karl W. Brennan,
et al., 606/228; 87/7; 428/224 [IMAGE AVAILABLE]

16.  4,470,941, Sep. 11, 1984, Preparation of composite surgical
sutures ; Leonard D. Kurtz, 264/136, 108, 134, 171, 174, 288.8, 290.5,
345; 606/230

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L7   (   1043648)S S
L8             3 S L5 AND L6
L9            20 S L6(3A)L1

=> d 13 37 57 188

37.  5,059,213, Oct. 22, 1991, Spiroid braided suture ; Michael P.
Chesterfield, et al., 606/228 [IMAGE AVAILABLE]

57.  5,019,093, May 28, 1991, Braided suture ; Donald S. Kaplan,
et al., 606/228, 230 [IMAGE AVAILABLE]

188.  4,470,941, Sep. 11, 1984, Preparation of composite surgical
sutures ; Leonard D. Kurtz, 264/136, 108, 134, 171, 174, 288.8, 290.5,
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L12            6 S L10 AND L7
L13           13 S L2 AND L4 AND L6 AND L7

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d 19 1 2 3 4 5

1. 4,470,941, Sep. 11, 1984, Preparation of composite surgical sutures; Leonard D. Kurtz, 264/136, 108, 134, 171, 174, 288.8, 290.5, 345; 606/230

2. 4,461,298, Jul. 24, 1984, Composite sutures of silk and hydrophobic thermoplastic elastomers; Shalaby W. Shalaby, et al., 606/231; 528/296

3. 4,441,496, Apr. 10, 1984, Copolymers of p-dioxanone and 2,5-morpholinediones and surgical devices formed therefrom having accelerated absorption characteristics; Shalaby W. Shalaby, et al., 606/230; 528/354; 606/231

4. 4,137,921, Feb. 6, 1979, Addition copolymers of lactide and glycolide and method of preparation; Yuzi Okuzumi, et al., 606/230; 525/411, 420; 528/354; 606/231; 623/1

5. 4,052,988, Oct. 11, 1977, Synthetic absorbable surgical devices of poly-dioxanone; Namassivaya Doddi, et al., 606/231; 528/354; 623/66

=> d 1 2 4 8 13

1. 5,147,400, Sep. 15, 1992, Connective tissue prosthesis; Donald S. Kaplan, et al., 623/13, 1, 11, 66 [IMAGE AVAILABLE]

2. 5,116,360, May 26, 1992, Mesh composite graft; Leonard Pinchuk, et al., 623/1, 11, 12 [IMAGE AVAILABLE]

4. 4,990,158, Feb. 5, 1991, Synthetic semiabsorbable tubular prosthesis; Donald S. Kaplan, et al., 623/1; 57/225 [IMAGE AVAILABLE]

8. 4,470,941, Sep. 11, 1984, Preparation of composite surgical sutures; Leonard D. Kurtz, 264/136, 108, 134, 171, 174, 288.8, 290.5, 345; 606/230

13. 3,748,828, Jul. 31, 1973, PROCESS AND APPARATUS FOR FLUID-LIQUID CONTACTING; Simon Lefebvre, 55/2, 29, 70, 73, 90, 93, 122, 233, 240, 300, 481, 527; 261/95, 103

=>

623 4-4C04

57 4-5D10

606 4-4C04

PATENT APPLICATION FEE DETERMINATION RECORD

Effective December 16, 1991

Application or Docket Number: 838511

CLAIMS AS FILED - PART I

	(Column 1) NUMBER FILED	(Column 2) NUMBER EXTRA	SMALL ENTITY RATE	SMALL ENTITY FEE	OR	OTHER THAN SMALL ENTITY RATE	OTHER THAN SMALL ENTITY FEE
FOR							
BASIC FEE	[redacted]		[redacted]	$ 345.00	OR	[redacted]	$ 690.00
TOTAL CLAIMS	24 minus 20 =	* 4	x $10 =		OR	x $20 = 80	
INDEPENDENT CLAIMS	1 minus 3 =	* 0	x 36 =		OR	x 72 =	
MULTIPLE DEPENDENT CLAIM PRESENT			+ 110 =		OR	+ 220 =	
			TOTAL		OR	TOTAL	770

* If the difference in column 1 is less than zero, enter "0" in column 2

CLAIMS AS AMENDED - PART II

		(Column 1) CLAIMS REMAINING AFTER AMENDMENT		(Column 2) HIGHEST NUMBER PREVIOUSLY PAID FOR	(Column 3) PRESENT EXTRA
AMENDMENT A	[redacted]		[redacted]		
	Total	* 24	Minus	** 24	=
	Independent	* 2	Minus	*** 3	=
	FIRST PRESENTATION OF MULTIPLE DEPENDENT CLAIM				
AMENDMENT B	[redacted]		[redacted]		
	Total	*	Minus	**	=
	Independent	*	Minus	***	=
	FIRST PRESENTATION OF MULTIPLE DEPENDENT CLAIM				
AMENDMENT C	[redacted]		[redacted]		
	Total	*	Minus	**	=
	Independent	*	Minus	***	=
	FIRST PRESENTATION OF MULTIPLE DEPENDENT CLAIM				

	SMALL ENTITY RATE	SMALL ENTITY ADDITIONAL FEE	OR	OTHER THAN SMALL ENTITY RATE	OTHER THAN SMALL ENTITY ADDITIONAL FEE
AMENDMENT A	x $10 =		OR	x $20 =	
	x 36 =		OR	x 72 =	
	+ 110 =		OR	+ 220 =	
	TOTAL ADDIT. FEE		OR	TOTAL ADDIT. FEE	
AMENDMENT B	x $10 =		OR	x $20 =	
	x 36 =		OR	x 72 =	
	+ 110 =		OR	+ 220 =	
	TOTAL ADDIT. FEE		OR	TOTAL ADDIT. FEE	
AMENDMENT C	x $10 =		OR	x $20 =	-
	x 36 =		OR	x 72 =	
	+ 110 =		OR	+ 220 =	
	TOTAL ADDIT. FEE		OR	TOTAL ADDIT. FEE	

* If the entry in column 1 is less than the entry in column 2, write "0" in column 3.
** If the "Highest Number Previously Paid For" IN THIS SPACE is less than 20, enter "20".
*** If the "Highest Number Previously Paid For" IN THIS SPACE is less than 3, enter "3".
The "Highest Number Previously Paid For" (Total or Independent) is the highest number found in the appropriate box in column 1.

Form PTO-875
(Rev. 12-91)

Patent and Trademark Office; U.S. DEPARTMENT OF COMMERCE

DePuy Mitek, Inc. v. Arthrex, Inc.
C.A. No.04-12457 PBS
DMI000330

STAPLE AREA

U.S. GOVERNMENT PRINTING OFFICE: 1991-294-566

PATENT NUMBER

APPLICATION SERIAL NUMBER
07/838,511

APPLICANT'S NAME (PLEASE PRINT)
Hunter et al.

IF REISSUE, ORIGINAL PATENT NUMBER

INTERNATIONAL CLASSIFICATION
A61F 1/00

ORIGINAL CLASSIFICATION

CLASS	SUBCLASS
606	231

CROSS REFERENCE(S)

CLASS	SUBCLASS (ONE SUBCLASS PER BLOCK)			
606	228			
87	7	9		
428	[illegible]	370		

GROUP ART UNIT: 1504

ASSISTANT EXAMINER (PLEASE STAMP OR PRINT FULL NAME)
CHRISTOPHER W. RAIMUND

PRIMARY EXAMINER (PLEASE STAMP OR PRINT FULL NAME)
GEORGE F. LESMES

PTO 270
(REV. 5-91)

ISSUE CLASSIFICATION SLIP

U S DEPARTMENT OF COMMERCE
PATENT AND TRADEMARK OFFICE

DePuy Mitek, Inc. v. Arthrex, Inc.
C.A. No.04-12457 PBS
DMI000331

PTO 1130 (REV 11-91)

U.S. DEPARTMENT OF COMMERCE– PATENT & TRADEMARK OFFICE

PACE DATA ENTRY CODING SHEET

1ST EXAMINER	ESTEY	DATE	3-4-92
2ND EXAMINER		DATE	

APPLICATION NUMBER	TYPE APPL	FILING DATE MONTH	DAY	YEAR	SPECIAL HANDLING	GROUP ART UNIT	CLASS	SHEETS OF DRAWING
07 838511	1	02	19	92		1304	264	3

TOTAL CLAIMS	INDEPENDENT CLAIMS	SMALL ENTITY?	FILING FEE	FOREIGN LICENSE	ATTORNEY DOCKET NUMBER
24	1	0	770	Y	ETH-782

CONTINUITY DATA

CONTINUITY CODE	STATUS CODE	PARENT APPPLICATION SERIAL NUMBER	PARENT PATENT NUMBER	PARENT FILING DATE MONTH	DAY	YEAR
		0				
		0				
		0				
		0				
		0				

PCT/FOREIGN APPLICATION DATA

FOREIGN PRIORITY CLAIMED	COUNTRY CODE	PCT/FOREIGN APPLICATION SERIAL NUMBER	FOREIGN FILING DATE MONTH	DAY	YEAR

DePuy Mitek, Inc. v. Arthrex, Inc.
C.A. No.04-12457 PBS
DM1000332

SEARCHED

Class	Sub.	Date	Exmr.
606	228	6/25/92	CWR
606	230	6/25/92	CWR
606	231	6/25/92	CWR
606	update 228	8/26/92	CWR
	230	↓	↓
↓	231		
606	228	11/9/93	CWR
↓	231	↓	↓
87	7		
↓	8		
	9		
428	225		

INTERFERENCE SEARCHED

Class	Sub.	Date	Exmr.
606	231	11/15/93	CWR

SEARCH NOTES

	Date	Exmr.
APS SEARCH	6/25/92	CWR
APS SEARCH	10/22/92	CWR

DePuy Mitek, Inc. v. Arthrex, Inc.
C.A. No.04-12457 PBS
DMI000333

Staple Issue Slip Here

POSITION	INIT.	DATE
CLASSIFIER	[illegible]	3-2-92
EXAMINER	[illegible]	3-4-92
TYPIST	[illegible]	3-5-92
CORPS CORR.		
SPEC. HAND.		
FILE MAINT.		

INDEX OF CLAIMS

Claim Final	Claim Original	Date 3/5/92	Date 11/2/92	Date 2/18/93	Date 8/15/93
1	(1)	+	—	—	—
2	2	+			
3	3	+			
4	4	+			
5	5	+			
6	6	+			
7	7	+			
8	8	+			
9	9	+			
10	10	+			
11	11	+			
12	12	+			
13	13	+			
14	14	+			
15	15	+			
16	16	+			
17	17	+			
18	18	+			
19	19	+			
20	20	+	↓	↓	↓
21	(21)	✓	✓	✓	=
22	22	✓	✓	✓	=
23	23	✓	✓	✓	=
24	24	✓	✓	✓	=
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SYMBOLS
- + ... Rejected
- = ... Allowed
- (Through oversight) ..... Canceled
- N ..... Withdrawn
- I ..... Interference
- A ..... Appeal
- O ..... Objected

Claim Final	Claim Original	Date
51		
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7-085

[illegible] 835511

[redacted]

APPROVED FOR LICENSE ☐

INITIALS MAR 0 3 92 4 6

RECEIVED
Received or Mailed MAR 0 9 1992
GROUP 150

CONTENTS

Entered or Counted			Received or Mailed
	1.	Application papers.	2/19/92
	2.	Prior Art	2/28/92
	3.	Rej. 3 mo.	8/11/92
	4.	Request for Recon	12/03/92
	5.	Rej. 3 mo.	12/11/92
	6.	Request for Recon	3-[illegible]
	7.	Rej. 3 mo.	8/9/93
	8.	[illegible]	8/9/93
	9.	[illegible]	8/9/93
	10.	[illegible]	8/9/93
	11.	[illegible]	11/18/93
	12.	[illegible]	[illegible]
	13.	[illegible] 148	[illegible]
	14.	3/8/[illegible] PTO GRANT MAY 24 1994	5/27/94
	15.	Notice of Allowance D 3311	
	16.		
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DePuy Mitek, Inc. v. Arthrex, Inc.
C.A. No.04-12457 PBS
DMI000335

A COVERING COVER 1-800-366-6060
5/8" Classic DTC 16834 for 32" Rivers

PTO-1683
(Rev. 7-96)